# Can we use wearable microtechnology devices to monitor motor changes post-concussion in team sport players?

# Can we use wearable microtechnology devices to monitor motor changes post-concussion in team sport players?

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This thesis is submitted in accordance with the requirements of The Graduate Research School, Australian Catholic University for the degree of Doctor of Philosophy by Laura Dunne

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# <u>Declaration</u>

*This thesis contains no material that has been extracted in whole or in part from a thesis that I have submitted towards the award of any other degree or diploma in any other tertiary institution. No other person's work has been used without due acknowledgment in the main text of the thesis. All research procedures reported in the thesis received the approval of the relevant Ethics/Safety Committees (where required).*

[redacted]

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# Published Works by the Author Incorporated into the Thesis

The studies outlined below were conducted during this PhD and are incorporated into the thesis. Each paper has been published following peer review.

1. Dunne, L.A.M., Cole, M.H., Cormack, S.J. *et al.* Validity and Reliability of Methods to Assess Movement Deficiencies Following Concussion: A COSMIN Systematic Review. *Sports Med - Open* **9**, 76 (2023). https://doi.org/10.1186/s40798-023-00625-0

**Statement of Authorship:**

Laura Dunne, Michael Cole, Stuart Cormack, David Howell, and Rich Johnston were involved in 60%, 19%, 7%, 7%, and 7% respectively of writing the systematic review.

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# Abstract

Sports-related concussions (SRCs) are common amongst athletes partaking in both contact (e.g., tackle) and non-contact (e.g., struck by ball) sports. Incidence of SRC has seen a steady increase amongst athletes, with proposed reasons including increased physicality, improved diagnostic tests, and increased awareness surrounding concussion and future complications of un-reported cases.

Concussion, otherwise referred to as mild traumatic brain injury (mTBI) is described as a transient disturbance of brain function due to direct (e.g., head contact) or indirect (e.g., whiplash) mechanisms. Due to the biomechanical properties of the brain, it has an increased susceptibility to injury under rotational accelerations typically associated with indirect impacts to the head. Symptoms following injury include headache, nausea, behavioural abnormalities (e.g., mood changes), vision and coordination deficits.

Current management of SRC is heavily reliant on subjective assessment tools, such as the Sports Concussion Assessment Tool (SCAT6) and, the recently introduced, office-based assessment (SCOAT6). These assessments are multidimensional to account for the wide variety of symptoms that may be experienced by individuals. They are used to assist with both diagnosis and return to play (RTP) clearance. Of particular interest is the assessment of motor function. The Balance Error Scoring System (BESS), it's modified version (mBESS), and tandem gait tasks are commonly used to assess motor performance, ultimately informing clearance for RTP. Of concern is the use of such tools, initially designed to assist with diagnosis, to evaluate RTP readiness. It is proposed that these assessments may lack the resolution or sensitivity to detect potential changes in motor function due to their simplicity and subjective nature. Given that there appears to be an increased risk of subsequent concussion and musculoskeletal injury upon RTP following SRC, players may be returning with lingering, subclinical deficits in function that go undetected by clinical tests. Laboratory assessments, such as 3D motion analysis or pressure sensitive walkways, may provide detailed insight into the deficits seen post-concussion; however, they are difficult to implement for most practitioners.

The aim of this thesis was to establish a more comprehensive motor assessment to further complement the current clinical assessments implemented as part of concussion management protocols. (1) Consolidate the reliability and validity of motor function assessments currently used in concussion management and further summarise their feasibility for clinicians and other end-users. (2) Assess the reliability of ankle, lumbar, and thoracic mounted devices to measure gait parameters on grassed surfaces. (3) Determine if lumbar- and thoracic-mounted IMUs can detect changes in movement strategies amongst athletes returning to play following a diagnosed SRC across a range of movement velocities. The findings from the systematic review identified limitations with currently implemented motor assessments; notably poor sensitivity of these assessments beyond the acute stages of concussion (>7 days). Most motor assessments rely heavily on subjective interpretation (e.g. number of errors) or blunt objective measures (e.g. time to complete task). They do not allow practitioners to understand what may be happening at a neuromuscular level to implement targeted rehabilitation strategies. The systematic review concluded that static and dynamic balance assessments may lack the requisite sensitivity to detect underlying neuromuscular changes in motor performance, especially when performed beyond acute stages. Gait-based assessments (single- and dual-task) were better at distinguishing between concussed and healthy populations, with instrumented versions being superior. Laboratory assessments are valid and reliable and are excellent instruments to obtain objective measures within a research setting. However, the accessibility, associated costs, and controlled environments mean they cannot be used by practitioners to monitor athletes returning to play. Thoracic worn IMUs are currently utilised by many team sport athletes to monitor training and game-play loads. The portability and low-cost of these devices make them suitable for team sport athletes to use within any desired environment. These devices have been shown to be effective at differentiating athletes suffering neuromuscular fatigue and it was hypothesised that this use may also be extended to monitoring the movement changes that present post-concussion.

To establish if IMUs can be used in the field to assist with concussion management, this thesis first sought to determine the test-retest reliability of ankle-, lumbar-, and thoracic-mounted IMUs to measure commonly reported linear and non-linear gait variables across set- and self-paced activities of increasing velocities performed on grassed surfaces. These IMU locations have previously demonstrated acceptable reliability for a range of outcome measures. However, these have only been performed within controlled, stable environments (e.g. running track, indoors). Access to these environments may be limited for many team sport athletes, therefore determining the reliability of these device locations on grassed surfaces was warranted.

Our findings showed that ankle-, lumbar-, and thoracic-mounted IMUs, in general, displayed acceptable test-retest reliability for a range of linear and non-linear gait variables. These results were consistent across velocities (walk, jog, run) and pace conditions (set- and self-paced). These findings support the use of IMUs to monitor common gait variables within the field, increasing feasibility for practitioners. Nonetheless, limitations for ankle- and lumbar-mounted IMUs do exist. Notably, these locations may impede activities performed in training and game environments (e.g. tackling, kicking a ball). Thoracic-mounted IMUs housed in a tight-fitted vest are already utilised by many team sport athletes and are less affected by sport-specific requirements, making them suitable for most athletes and sports. Practitioners can be confident that thoracic-mounted IMUs can reliably measure variables of gait over a five-week period based on the current research.

Upon establishing the test-retest reliability of ankle-, lumbar-, and thoracic-mounted IMUs, this thesis sought to determine whether there were differences in gait derived from the lumbar- and thoracic-mounted IMUs between concussed and control athletes at a range of movement velocities. To date, post-concussion gait assessments have been isolated to walking tasks which may not accurately represent the demands required upon return to play. Therefore, we aimed to investigate if increasing velocities (jog and run) also displayed between-group differences and assess whether they existed beyond RTP clearance.

Our findings suggest that variations in gait strategies still exist beyond RTP timelines. These variations were evident across linear and non-linear outcome measures collected via lumbar- and thoracic-mounted IMUs. Previous literature has suggested concussed athletes have a slower gait when walking. Whilst our findings supported this, concussed athletes performed jog and run activities at faster speeds than healthy controls. Largest between-group differences were seen for measures of sample entropy (vertical and mediolateral acceleration), mediolateral acceleration profiles, and centre of mass (COM) displacement.

Concussed athletes demonstrated greater mediolateral movement, indicating more side-to-side movement, and reduced COM displacement and vertical acceleration, suggesting reduced lower-limb stiffness. Non-linear analysis showed greater irregularity for vertical accelerations across all activities for concussed athletes, indicating a less predictable movement pattern for these participants. In general, this score improved for walk and run activities, with no change seen for jogging.

Differences between device locations were particularly evident for the assessment of sample entropy and mediolateral acceleration. Data obtained from the thoracic-mounted IMUs suggest greater variability in mediolateral acceleration, while the lumbar-mounted IMUs only showed greater irregularity for walk activities. Due to being further up the kinetic chain, thoracic-mounted IMUs may be more influenced by upper-limb and trunk movements, whereas measures from lumbar-mounted IMUs may better represent lumbo-pelvic movements and COM displacement. Both sites can provide valuable information, but practitioners must be aware of the limitations or bias (e.g. magnitude of some outcome variables) associated with each location when interpreting results. Notably, lumbar-mounted devices are not commonly worn and therefore require the attachment of an additional device. Thoracic mounted devices are commonly worn during team sport training and competition and may allow for pre-injury measures and regular post-injury measures of gait to be performed with relative ease to aid decision making in team sport.

In conclusion, the studies within this thesis found that ankle-, lumbar-, and thoracic-mounted IMUs can reliably measure a range of linear and non-linear gait metrics during activities across a range of velocities when performed on grassed surfaces. In addition, lumbar- and thoracic-mounted IMUs can be used to show differences between concussed and healthy athletes for various linear and non-linear outcome measures up to 50-days post-concussion.

These findings support the use of IMUs in the field to provide a more comprehensive assessment of gait following concussion. The low-cost and portability of IMUs make them easy for many team sport athletes and practitioners to access and use to assess any potential changes in gait strategies following injury. Use of instrumented gait assessments across increasing velocities may provide important insight to inform an athlete's RTP readiness when compared with more traditional clinical assessments (e.g. BESS) and may inform practitioners of additional assessment requirements to ensure player safety and well-being following RTP.

# Summary


In summary, the findings of this thesis demonstrate that:

1. Subjective assessments (e.g., BESS, mBESS, tandem gait) used to assist with RTP clearance display poor reliability and/or validity to detect changes in motor strategies beyond acute (>7 days) stages of concussion.
2. Instrumenting these assessments through the use of laboratory equipment (e.g., force plates) or wearable technology (e.g., IMUs) improves their reliability and sensitivity. However, gait assessments may still provide greater insight into potential changes in motor strategies post-concussion.
3. If logistics allow, clinicians and practitioners should aim to implement instrumented gait assessments where possible as opposed to instrumented or non-instrumented static or dynamic assessments.
4. Ankle-, lumbar-, and thoracic-mounted units can reliably measure linear and non-linear gait metrics on grassed surfaces over increasing velocities.
5. Lumbar- and thoracic-mounted IMUs are able to detect differences in gait amongst athletes returning from a SRC. Notably, concussed athletes demonstrate reduced regularity of gait and reductions in vertical accelerations or vertical ground reaction force (GRF). This seems to be exacerbated when velocity increases. Differences in gait can be seen >50-days post-concussion.
6. Thoracic-mounted IMUs may provide a more comprehensive analysis of gait following concussion when compared to current clinical assessments. IMUs and GNSS devices are a cost-effective and feasible option for team sport athletes to collect data within the field without disturbing their training schedules. The information derived from these devices can inform practitioners if further investigation is required and guide their assessment of a player’s return to play (RTP) readiness.

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# List of Abbreviations

All abbreviations are presented in full upon first use in each chapter or study before appearing abbreviated. Below provides a list of all abbreviations.

| | |
|---|---|
| AMPA | a-amino-3-hydroxy-5-methyl-4-isoxazolepropionic acid |
| ARAS | Ascending reticular activating system |
| ATP | Adenosine triphosphate |
| AUC | Area under the curve |
| BAM | Balance accelerometry measure |
| BBB | Blood brain barrier |
| BESS | Balance error scoring system |
| BESTest | Balance Evaluations System Test |
| CAREN | Computer-Assisted Rehabilitation ENvironment System |
| CHII | Cumulative head impact index |
| ClinROMs | Clinical reported outcome measures |
| CNS | Central nervous system |
| COM | Centre of mass |
| COP | Centre of pressure |
| COSMIN | Consensus-based Standards for the selection of health Measurement Instruments |
| CSP | Cortical silent period |
| CTE | Chronic traumatic encephalopathy |
| DPSI | Dynamic postural stability index |
| DT | Dual-task |
| GABA | Gamma-aminobutyric acid |
| GNSS | Global navigation satellite system |
| GRF | Ground reaction force |
| GRTP | Graduated return to play |
| HAE | Head acceleration event |
| HDOP | Horizontal Dilution of Precision |
| HIE | Head impact exposure |
| ICC | Intraclass correlation coefficient |
| IMU | Inertial measurement unit |
| KPR | Kash pulse recovery |
| Kvert | Vertical stiffness |
| LOS | Limits of stability |
| mBESS | Modified balance error scoring system |
| MCTSIB | Modified clinical test of sensory interaction in balance |
| MDC | Minimal detectable change |
| mTBI | Mild traumatic brain injury |
| NMDA | N-methyl-D-aspartate |
| PANESS | Physical and neurological examination of subtle signs |
| PCSS | Post-concussion symptom scale |
| PCTSIB | Paediatric clinical test of sensory interaction in balance |
| PPCS | Persistent post-concussion symptoms |
| PRISMA | Preferred Reporting Items for Systematic Reviews and Meta Analyses |
| PST | Postural stress test |
| PVGRF | Peak vertical ground reaction force |

| | |
|---|---|
| RTP | Return to play |
| SCAT | Sport concussion assessment tool |
| SCOAT | Sport concussion office assessment tool |
| SEM | Systematic error of the measure |
| Sens | Sensitivity |
| SIS | Second impact syndrome |
| SLS | Single leg stance |
| SOT | Sensory organisation test |
| Spec | Specificity |
| SRC | Sports-related concussion |
| TUG | Timed up and go |
| YBT | Y-balance test |

# Chapter 1

# General Introduction

## 1.1 Scope of thesis

Concussion, otherwise known as a mild traumatic brain injury (mTBI), can be described as a transient disturbance of brain function (1) resulting from direct (e.g. struck on the head), or indirect (e.g. whiplash mechanism) trauma to the head (2). This thesis primarily focused on sports-related concussions (SRC) occurring amongst team sport athletes. Notably, this thesis aimed to describe the mechanistic properties of concussion, incidence amongst team sport athletes, and the implications of current management strategies to gauge return to play (RTP) readiness.

Research has identified a potential association between concussion and an increased risk of subsequent injury upon clearance to RTP (3-6). Deficits in gait, postural control, reaction time, and cognition are suspected contributors to this increased risk (3, 5-9). Current assessment of motor performance (static, dynamic, and gait) in athlete populations is heavily reliant on subjective field-based tools (i.e. Balance Error Scoring System (BESS), timed tandem gait) due to their convenience and ease of interpretation. However, it is suspected that these tools may lack the requisite sensitivity to detect more subtle variations in motor strategies, potentially prematurely clearing athletes for RTP. To understand the association between concussion and injury risk, readers must first develop a sound understanding of the mechanistic properties of concussion as well as the current tools used to assess motor strategies.

## 1.2 Overview of concussion

### 1.2.1 Incidence

Due to a lack of agreement surrounding diagnostic criteria, as well as the multitude of causal mechanisms, true incidence of concussion may be underestimated (10). In the United States alone, 1.6 to 3.8 million concussions are estimated to occur annually amongst the general population (6, 11, 12). More specifically, approximately 16% of children aged ten years or younger (13) and 20% of adolescents (14) have sustained at least one diagnosed concussion requiring medical attention.

Within sport, the widespread incidence of SRC is a public health concern given a significant proportion of society participates in regular sport (15). Insights from the 2020 AusPlay Survey (16), a national Australian survey tracking sports and physical activity participation across Australia, indicated that approximately 83.3% of women and 79.4% of men participated in some form of physical activity each week, with 48% of children participating in organised sport or physical activity outside of school (16). It is suggested that approximately 20% of total reported concussions are due to sports participation (17). An observational study of concussion incidence in New Zealand indicated approximately 16,994 SRC were reported between 2018 and 2019 (18), this accounted for approximately 48% of all reported concussions and traumatic brain injuries during that period (18). Similarly, match injury surveillance of English professional rugby union matches showed concussions to be the most common injury across the 2011 to 2019 seasons (19). Clearly, there are a great number of sports players who have sustained or are at risk of sustaining a concussion. Therefore, understanding how to appropriately return these players to sport is vital.

Athlete exposure or player hours (e.g., per 1000 training hours) is often used to calculate and measure injury incidence (17, 18). With regards to SRC, these methods provide data on the frequency of these injuries relative to match play and training hours (15, 18). Contact sports with a greater number of collisions commonly present with the highest rates of SRC (15, 17, 18). Incidence of concussion reporting has increased over the years. In rugby union players incidence and severity (number of days absent from full training or match-play) increased from 4.3 per 1000 match-hours in the 2002/03 season to 20.9 per 1000 in the 2016/17 season (20). Severity of concussion, measured as the mean and median number of days absent from training and/or match play, significantly increased after the introduction of the graduated return to play (GRTP) guidelines in 2011 (20). This is likely due to a reduction in the number of players returning within six days post-concussion (20). Concussion and injury to the head and face displayed greatest incidence compared to all other reported injuries over the 16-year review (20). Increased awareness and knowledge surrounding concussion may contribute to the rise in incidence observed (21). Rugby league has also shown high rates of concussion incidence, with approximately 14.7 concussions per 1000 playing hours recorded during the 2019 season (22, 23). Pooled incidence between rugby league and rugby union is estimated to be between 8 and 28 concussions per 1000 match hours for males (15). Women's professional rugby league has demonstrated similar trends with an estimated 33 concussions per 1000 match hours (24). Estimated concussion incidence is 6.1-10.3 per 1000 match hours with an average of 33-days lost (25, 26). At the youth level, an 11 year observation of high school SRC incidence in the United States saw a 15.5% increase across the study (21). Increased awareness of the risks and improved detection protocols may have largely contributed to this rise amongst adult and youth athletes (21); although increased physical performance and competitiveness of teams may also influence incidence (15).

Concussion incidence is highly dependent on the sport, playing position, competition level, and physiological and anthropometric characteristics (15, 27). Differences in mechanisms are evident across a variety of team sports. For example, soccer players are more likely to sustain a concussion via heading the ball, or player-to-player contact when heading the ball (28, 29), while other football codes, such as Australian football, gridiron (American football), and rugby (league and union), report player-to-player contact during tackling as the most common mechanism (18, 28). Specifically, contact of the head with an opposition's upper limb (i.e., above sternum) or lower limb (i.e., knee) were found to be the most common causes of concussion in rugby (league and union) and Australian football (18, 30). In gridiron and ice hockey, a direct hit to the head was the primary cause (18), while tackling was identified as the most common cause of concussion in rugby league (25, 30). Differences in concussion mechanisms between sexes have been shown in rugby league. For male athletes, the highest concussion risk is associated with being the tackler compared to the ball carrier (31, 32). Whereas, in female rugby league athletes one study has reported a greater incidence for the tackler (30) and another showing no difference between roles (24). Speculatively, these differences between and within sexes may be due to the contact proficiency and training ages of male and female rugby league players, as well as the differences that are likely between elite female competitions played in England and Australia.

There are other factors that also influence concussion rates. Game intensity, number of accumulated plays, and game duration are considered potential contributors to concussion incidence (33). Notably, increased incidence may be seen during games of greater importance (e.g. play-offs compared to regular in-season games) (34) or longer duration games (35). It is suggested that player fatigue may increase susceptibility of sustaining a concussion (35). Positional differences within specific sports also influence risk. In rugby union, half-backs were identified as being at greatest risk of concussion. It is suspected that the lower body mass of these players may be a contributing factor (36). Taller and lighter players have shown greater susceptibility to sustaining a concussion during game-play irrespective of position (36). Consistent across the literature, head-neck strength was seen as a modifiable risk factor, with research suggesting those with lower strength were more susceptible to greater inertial forces upon head contact (18, 28).

## 1.3 Injury mechanism and pathophysiology

### 1.3.1 Biomechanical mechanism of injury

The brain is situated within the skull surrounded by cerebrospinal fluid (CSF) and is one of the most complex systems within the human body (37). The brain consists of four main regions; cerebrum, cerebellum, brain stem, and diencephalon (37). The cerebrum, which includes the cerebral cortex, accounts for majority of the brain's volume and hosts a large portion of neurons to assist with transmitting and receiving information for higher functioning (37, 38). The neurons (and their axons) within this region are located in grey and white matter. This interface is believed to be the primary site of axonal injury during concussion (39). The cerebellum, brain stem, and diencephalon all contribute to neural signalling processes to maintain whole body function, including motor, regulatory, and vital functions (37, 38).

#### 1.3.1.1 Mechanical properties of the brain

When considering injury mechanisms of the brain, we must first consider the mechanical properties of the tissue (40, 41). The human brain is a soft tissue suspended in CSF, encased by the skull (42, 43). Mechanical loads that occur through everyday activities (i.e. changes in posture) are well tolerated due to the ability of the brain to oscillate within the CSF (42). However, the brain lacks the ability to adapt to mechanical strain and accelerations, which can result in structural injury and/or functional impairment (39, 42). The brain is comprised of grey and white matter. White matter contains myelinated axons, whilst grey matter consists of unmyelinated axons and cells bodies (39, 42). Differences in mechanical properties of grey and white matter exist, with their different viscoelastic properties suggesting that white matter is stiffer and less adaptable to strain (39). This difference seems to be rate dependent, with increased stiffness occurring at increased strain rates (39, 42), implying a reduced ability to withstand deformation at greater rates of tissue acceleration. When considering location, increased axonal damage is believed to occur at the grey-white matter interface. Grey matter consists of cell bodies, dendrites, and unmyelinated axons whereas white matter primarily consists of myelinated axons (44). Although still poorly understood, variations in density as a result of structural differences are believed to contribute to the variances seen between grey and white matter stiffness (44). This may result in differences between tissue tolerance during head acceleration events (e.g. concussion) across both these regions. Due to the variation in strain tolerance and density between the two, the degree and rate of deformation would be different (39). This may result in varying amounts of shear deformation occurring along different aspects of the axon and disproportionate movement between the structures (39).

#### 1.3.1.2 Mechanical origins of concussion

Concussion is characterised by a near instant transfer of kinetic energy that is quickly expended (45). Unlike more severe traumatic brain injuries (TBI) visible structural damage to the brain is largely absent using traditional neuroimaging practices (40, 41). Concussion may occur from direct (i.e. struck on head) or indirect (i.e. whiplash mechanism) trauma to the head, which produce inertial accelerations (41). These inertial accelerations can be further categorised into linear or rotational (41). Linear accelerations refer to a change in the velocity of the head along the sagittal plane (e.g., in a forward to backward direction). Early investigation of concussion concentrated on linear accelerations resulting from head impacts (40). Recordings of pressure within brain models correlated with peak acceleration, identifying a relationship between the two (40, 41). Previously, it was believed that such accelerations would cause a coup-contrecoup injury (40, 41, 46). Implying that accelerations and decelerations would cause the brain to strike the inner skull. However, due to the absence of visible structural damage, the coup-contrecoup mechanism is not widely supported. Instead, literature points to a transient increase in pressure gradients creating strains, which lead to temporary neurologic dysfunction (40, 41). Rotational accelerations involve the head being accelerated around the body’s or body segment’s centre of gravity (45). The mechanical properties of the brain have relatively high susceptibility to shearing and tensile forces due to the increased stiffness and viscoelastic properties (2, 45, 47, 48). Therefore, rotational accelerations creating larger strains present a higher likelihood to result in concussion as well as cause loss of consciousness (41).

When the head experiences linear or rotational accelerations, the brain responds by oscillating, rotating, and gliding within the skull (45). However due to the nature of brain tissue and its suspension within CSF, the brain is easily distorted and is susceptible to forces acting externally on the body (45). Notably, shearing and stretching forces may result in microvascular damage and neural dysfunction within the brain (40, 41, 49). This is considered to play a large role in the causation of concussion (41).

#### 1.3.1.3 Injury thresholds

Three general categories describing mechanisms of injury risk have been proposed: (1) mechanical load that exceeds the mechanical tolerance of brain tissue, (2) inability to tolerate normal mechanical loads due to a reduced tolerance, (3) high mechanical load coupled with reduced tolerance (40). It is important to consider that these proposed mechanisms will directly affect the load and rate of force that may cause a concussion. Previous studies have sought to identify biomechanical impact thresholds, but the threshold estimates for linear accelerations vary significantly (50-53). Linear accelerations ranging from 66g to >90g have been correlated with the likelihood of sustaining a concussion (50-53). Estimations for rotational accelerations are thought to range from 4000 $rad{\cdot}s^{-2}$ to upwards of 9000 $rad{\cdot}s^{-2}$ (54-56). Due to the varying magnitudes, direction of impact (coronal, horizontal, sagittal planes), areas of impacts, characteristics of acceleration, lack of uniformity in measurement acquisition, and history of concussion or non-concussive impacts, it can be difficult to establish definitive thresholds for concussive events.

### 1.3.2 Acute concussion pathophysiology

#### 1.3.2.1 Energy loss

Upon the initial head acceleration, stretching of the axons can cause an efflux of potassium and influx of sodium and calcium, causing non-specific depolarisation (45, 48, 57-59). This leads to the release of excitatory neurotransmitters, such as glutamate which subsequently further increases depolarization via kainite, N-methyl-D-aspartate (NMDA), and α-amino-3-hydroxy-5-methyl-4-isoxazolepropionic acid (AMPA) receptors (57). In general, these receptors work to mediate excitation at the synaptic transmission and modulate ligand-gated ion channels (2, 45, 48, 57, 60-62). Increased activation leads to; (1) an influx of calcium causing mitochondrial dysfunction, significantly hindering its ability to create adenosine triphosphate (ATP) via aerobic metabolism (57), and (2) dysregulation of the sodium potassium pump due to potassium efflux (57). This mass depolarisation presents as the immediate signs and symptoms of a concussion, which include a loss of balance or consciousness, seizures and tonic posturing.

Additionally, small disruptions in the vasculature may disrupt the integrity of the blood-brain barrier (BBB), increasing its permeability (63). This occurs despite any structural damage being visible on imaging (63). Studies have shown that this disruption can also occur independent of axonal injury (63). Increased permeability may then result in cerebral oedema as well as alterations to the microenvironment, causing subsequent disruptions within other processes (57).

#### 1.3.2.2 Energy crisis

To restore homeostasis and resting membrane potential, the sodium-potassium pump activity increases, requiring large quantities of ATP (48, 57). However, due to the reduced capacity to produce ATP via aerobic metabolism, there is an increase in glycolysis, a form of anaerobic metabolism (48, 57). Subsequently this mechanism produces substantially less ATP (net value of 2 mol ATP per mol glucose, compared with a net value of 32 mol ATP per mol glucose via aerobic metabolism) and produces lactate as a by-product, further altering the microenvironment. Over time, the increased demand for ATP cannot be met, leading to an "energy crisis", that underpins the physical symptoms of concussion (e.g. headache, nausea, motor deficits) (45, 48, 57).

It is proposed that this neurometabolic cascade resulting in an energy crisis may cause deficits in processing speed, learning capabilities, attention, and executive functioning (7). Collectively these impairments are seen as primary causes of reductions in focus and task performance following concussion (48, 64). Additionally, impairment of the visual and motor pathways leads to dysfunction of the vestibulo-ocular system, which also contributes to the symptoms of dizziness and vision disturbance that contribute to the balance impairments commonly observed (47, 64, 65). Severity of impact and subsequent damage to the brain heavily influence the impairments and recovery timelines experienced by each individual (2, 41).

### 1.3.3 Signs and symptoms

Concussion is a multifaceted injury affecting cognitive, physical, and psychological function (7). Due to the complexity surrounding the aetiology of this injury, presentation of signs and symptoms, along with recovery timeframes varies amongst individuals (2). Approximately 10% of concussions include loss of consciousness (66), with evidence suggesting rotational accelerations of the head and increased cerebral ischemia are primary contributors (41, 45, 67), Specifically it is hypothesised that due to the shear strain associated with rotational accelerations, the ascending reticular activating system (ARAS), which assists in maintaining consciousness, may be disrupted, resulting in momentary loss of consciousness (67).

Immediate symptoms following impact typically include headache, dizziness, and nausea (2). In addition, on-field assessments (e.g. SCAT6) assist in identifying cognitive, memory, and motor (e.g. static balance) deficits that are present during this acute stage (2). As time from injury increases, observable symptoms begin to dissipate. Clinical recovery is often observed within one month of injury, however individual characteristics and concussion severity may extend recovery times for selected individuals (2, 47). Individuals who sustained a concussion with loss of consciousness may report increased symptom duration due to the greater severity of the injury (68). Once clinically recovered, individuals are typically asymptomatic, however may still experience sleep disturbances or motor problems that are exacerbated during sports participation (69).

Symptoms following a concussion often fall into four broad categories; affective, cognitive, physical, and sleep (2) (Table 1.1). Affective alterations, typically present during the initial 7-10 days following a concussion, and include those associated with changes in mood or behaviour (2). Individuals may experience irritability or distinct mood changes during the acute stages of concussion (2, 70) and symptoms persisting beyond three months may indicate persistent post-concussion symptoms (PPCS) (47, 71). Exact explanations as to why these occur are mostly unknown, however it is hypothesised that a combination of metabolic disturbances, sleep abnormalities, emotional stress, and alterations in executive brain function may contribute to affective alterations (45, 48, 72).

Observable symptoms including cognitive deficits following a concussion are most pronounced in the first seven days , however are typically resolved within two to four weeks (3). Common signs and symptoms include; confusion, amnesia, mental fogginess, speech impairments, and/or difficulty concentrating (2, 70). It is thought that, in an attempt to restore homeostasis upon the metabolic cascade of events, a subsequent reduction in available energy within the brain may contribute to reductions in cognitive function (2).

Physical impairments include symptoms such as; headache, dizziness, nausea, migraines, visual, and motor impairments (2). With regard to motor impairments, individuals typically experience poor coordination, balance disturbances, and/or gait unsteadiness (2, 64). Physical symptoms are easily identified during acute stages via standard clinical assessments such as symptom scales (e.g., SCAT and PCSS) and static and dynamic balance assessments (e.g. BESS). However, lingering motor deficits following symptomatic and clinical recovery have been identified (4, 6, 73-75). Clinical assessments, such as the BESS, may lack the required fidelity to replicate return to sport demands. Therefore, athletes may pass these tests however still present with motor deficits upon RTP. Potential alterations in primary motor cortex function and intracortical inhibition may also prolong these deficits (6, 76).

Sleep disturbances including hypersomnia or insomnia may also be associated with acute stage concussion (2). Hypersomnia, characterised by the drive to sleep during the day, is most commonly reported, with individuals usually returning to normal sleeping patterns within 7 to 10 days (69). Insomnia, characterised by the inability to fall asleep, appears more likely following the initial hypersomnia phase during the acute stage of injury. It is suggested that neurometabolic cascade following initial injury may cause dysregulation of the sleep and arousal centres in the brain, thus leading to sleep abnormalities (77).

Table 1.1 Summary of general symptoms following sports-related concussion

| Category | Common symptoms | Average recovery | Proposed mechanisms |
|---|---|---|---|
| Affective | Alterations in mood/behaviour:<br>• Depressive symptoms<br>• ↑ irritability | 7-10 days<br>2-4 weeks | Sleep disturbances<br>• Day time sleepiness<br>• Sleep deprivation<br>Multi-dimensional disruption of limbic system leading to failure to maintain homeostatic emotional control<br>Metabolic disturbances<br>Increased emotional stress<br>Alterations in executive function |
| Cognitive | Confusion, amnesia, mental fogginess, speech impairments, decreased concentration | 7-10 days<br>2-4 weeks | Reductions in available energy due to an attempt to restore homeostasis |
| Physical (symptoms) | Headache, migraine, dizziness, nausea | 7-10 days | Metabolic disturbances<br>Reductions in available energy due to an attempt to restore homeostasis |
| Physical (motor) | ↓ coordination, ↓ balance (static and dynamic), gait abnormalities, ↓ maximal muscle force, ↓ lower limb stiffness | 7-10 days<br>2-4 weeks<br>>3-6 months | Decreased primary motor cortex function:<br>• ↓ metabolic activity<br>• ↑ intracortical inhibition<br>• ↓intra-cortical facilitation → Hypoexcitability<br>Vestibulo-ocular deficits<br>• ↓ near point convergence (NPC)<br>• Convergence insufficiency |
| Sleep | Hypersomnia (increased sleep)<br>Insomnia (difficulty sleeping) | 7-10 days | Dysregulation of sleep and arousal centres:<br>• ↓ melatonin due to damage to pineal gland<br>• ↓ hypocretin by hypothalamus<br>• Inconsistent or shifted sleep schedule |

Abnormal recovery timeframes characterised by persistent symptoms for three or more months may indicate the presence of post-concussion syndrome (PCS), more recently referred to as persistent post-concussion symptoms (PPCS). Approximately 10 to 30% of individuals will experience PPCS following a concussion (78). Lack of consensus regarding diagnostic criteria for this has produced uncertainty surrounding detection and diagnosis amongst health professionals (47, 69, 79, 80). The Diagnostic and Statistical Manual of Mental Health Disorders state that; for the presence of PPCS to be determined, an individual must meet specific pre-determined criteria. This includes evidence of cognitive deficits seen via a neurobehavioral assessment, and the patient must experience three or more of the following symptoms; headache, dizziness, irritability, personality changes, fatigue, sleep disturbance, or affective disturbance (69). Despite these clear criteria, the lack of specificity regarding the origin of these symptoms and the inability to determine that they are directly related to the concussive injury makes diagnosis difficult. Exact pathophysiology underlying PPCS is unknown, however it is hypothesised that prolonged neuroinflammation or alterations in cerebrovascular activity may impede recovery, and therefore manifest as PPCS (69). Recovery from PPCS includes correcting any sleep abnormalities in addition to a graduated return to activities of daily living and exercise (69).

Although rare, an additional complication that may follow concussion is second impact syndrome (SIS) (2). This is characterised by a subsequent injury to the head before complete neurological resolution (2, 41). Exact pathophysiology is not completely understood and, in some cases, the existence of SIS is questioned. It is proposed that a second impact my lead to increased oedema in the brain (81). When a subsequent force to the head occurs before resolution of acute disturbances it may result in a secondary release of inflammatory markers, thus causing rapid and severe swelling on the brain (2). If not appropriately managed, SIS can result in severe neurologic complications, and potentially death (2). This emphasises the need for 1) strict return to play (RTP) guidelines to minimise risk of severe outcomes related to subsequent impacts and 2) compliance to the recommended stand down periods (average of 11-days) (82) from both athletes and staff.

### 1.3.4 Head impact exposures and head accelerations events

Repetitive head impacts or accelerations are associated with playing contact sports (83, 84). Whilst some of these impacts lead to concussion, most do not (85). These head impacts have been quantified and are commonly referred to as head acceleration events (HAE) and can be aggregated as head impact exposure (HIE). These events (i.e. tackle, head contact with ground, head contact with object) are commonplace in most contact sports, however often go undetected due to a lack of obvious signs and symptoms typically associated with concussion (86, 87). It has been suggested that accumulation of these impacts may result in neurological impairments and potential reduction in head impact tolerance, meaning individuals may be more susceptible to sustaining a concussion under lower than normal magnitudes (40, 84, 85). Previous research has also identified the possibility of impaired blood-brain barrier permeability, axonal injury, and acute bouts of neuroinflammation occurring over periods of repeated impacts (65).

Studies investigating the clinical presentations of accumulated head impacts have demonstrated motor and cognitive deficits. With regard to cognitive deficits, declines in cognition (e.g. working memory) have been seen amongst high school and college football and ice hockey athletes' post-concussion (88-90). Of particular interest in relation to this thesis, are the deficits reported in motor tasks, which appear to last longer than other symptoms of a concussion. Notably, postural control deficits, specifically, increased sway velocities and increased centre of pressure (COP) area when assessed via instrumented tools (65) have been seen following head impacts in the absence of a diagnosed concussion. These findings are consistent with research investigating pre- and post-season changes in neuromuscular function amongst female varsity rugby union players (86). Although participants were not diagnosed with a concussion, deficits in bilateral stance on firm surface and tandem stance on a foam surface were evident in post-season assessments (86). It can be assumed that such deficits in static balance may be exacerbated during gait-based tasks due to the increased requirements of the CNS (91).

It is suggested that long-term complications of successive concussions or micro-trauma associated with repetitive head impacts may result in chronic traumatic encephalopathy (CTE)

(87, 92). This is characterised by progressive neurodegeneration of the brain and includes memory deficits, motor and behavioural abnormalities, and potential speech impairments (2, 87). It has been suggested that changes in white matter tracts throughout the course of repetitive head impacts may be an underlying cause of this condition (87). This is of particular concern due to the proportion of concussions that go undiagnosed (93) as well as the repeated head trauma that many professional team sport athletes are potentially exposed to throughout their sporting career (86). As such, it is vital that practitioners and clinicians have the appropriate diagnostic tests available to make the correct decisions regarding RTP.

Implementation of accelerometers and gyroscopes embedded in helmets (American football) and mouthguards have allowed the quantification of linear and rotational head accelerations during training and match play situations, which allows the number and magnitude of head impacts a player receives to be approximated. This can then be combined into a cumulative head impact index (CHII) (84, 92, 94-96). Rugby codes, amongst other contact sports, have consistently high injury rates due to the game's demands (26, 97). Specifically, the risk of head injuries during tackles may have long-term implications for a player's safety and wellbeing. Analysis of elite women's rugby league found 28% of tackle events involved contact with the head (30). While only a portion of these head impacts result in concussion, players are accumulating numerous non-concussive head impacts, potentially increasing their susceptibility to a reportable SRC event in the future (98). Although relatively new, findings show that quantifying cumulative linear and rotational head accelerations may be more predictive of CTE than the number of concussions sustained (92). Data from American football has demonstrated that each additional year of play was associated with a 15% increased risk of having a CTE diagnosis (92). Not only does this information assist with further investigations of CTE, but it can also be used to mitigate repeated head impacts by addressing game-play policy and rules to improve player safety.

### 1.3.5 Subsequent injury upon return to play

The effects of concussion on cognition and motor control have been extensively explored, with majority of research identifying a potential association between RTP and an increased risk of subsequent injury (3-6). More specifically, an increase in lower-limb musculoskeletal injury incidence amongst those returning from concussion has been observed (4-6, 99). These increases have been seen up to one-year post-injury (5, 6, 99). Notably, players returning from concussion have demonstrated a 2.5 times higher likelihood of sustaining a non-contact injury within 90 days of RTP (100). When pooling contact and non-contact injuries, the likelihood increases to approximately 3.4 times greater odds among previously concussed players (101). It is hypothesised that these athletes may be unable to effectively register external cues and/or actions of opponents due to a reduction in processing speed, reaction time, and/or cognition (101). This may impact a player's decision-making abilities during game-play. Studies examining the effects of concussion one year after injury found professional athletes and collegiate athletes were at a 2.2 times and 2.5 times greater risk of sustaining an injury, respectively (75). A 2019 study conducted on athletes returning from concussion summarised odds ratios for participants, indicating odds of 2.8 for the risk of a participant experiencing any injury; 4.4 for a second concussion; 1.8 for a lower extremity injury (102). Research has shown that an increase in the symptom-free waiting period (up to 13-days) prior to RTP clearance may reduce injury risk upon RTP, potentially providing more specificity around RTP guidelines for athletes (103). Studies that found no significant increased risk of injury upon RTP highlighted the potential benefits of having the same medical practitioner performing the diagnosis and implementing more comprehensive, multifactorial testing protocols prior to RTP (104, 105). It is suspected that these factors may (1) provide more consistency with concussion diagnosis and stand-down time frames and (2) better assess RTP readiness based on a range of factors required for game-play (e.g. decision making, motor function). However, despite these proposed improvements, the direct assessment of the underlying mechanisms affecting motor control are yet to be taken into consideration.

Motor deficits seen upon RTP from concussion have been centred around gait and postural control (3, 5-8). However, impairments in cognitive function and memory have also been found and should be considered within a multidisciplinary approach (106). The time course of these subclinical changes in function are unclear but can persist for up to a year post-concussion (6, 73, 99). Dual-task assessments seem to exacerbate cognitive and motor deficits, with concussed participants displaying increased cognitive errors during assessment (107). In regards to movement characteristics, conservative gait, typically characterised by a reduction in velocity, increased medio-lateral sway, and an increased time in double support (5, 6) has been seen 28 days post-concussion (108). It is hypothesised that a contributing factor to the typical gait characteristics seen post-concussion may be associated with poor sensorimotor control, highlighting a reduced capacity for the brain to appropriately process sensory information to produce coordinated movements within the environment (109). Additionally, impairments in termination of gait and propulsive and braking forces have also been observed amongst this population (5). It has been proposed that these alterations may increase during game-play, and therefore significantly contribute to injury risk upon RTP (5, 6, 108).

Post-concussive deficits are worsened during dual-task testing (7, 8, 108, 110, 111), where attention is divided between a cognitive and motor task. It has been argued that assessments integrating a dual-task approach may be more reflective of the team sport situations, due to concurrent motor and cognitive demands (7). Studies have shown significant deficits in either motor or cognitive tasks amongst those with a previous concussion (5, 6, 75, 102, 112). Lingering deficits in brain function, specifically executive functioning, may reduce an athlete's capacity to effectively divide attention between cognitive and motor tasks (110, 113-115). Due to this, characteristics seen during single-task assessments may be exacerbated with the addition of a cognitive task (116). These results may also play a role in understanding subsequent injury following a concussion, and more specifically motor deficits that tend to go undetected during subjective clinical assessments. Despite these deficits in neuromuscular control that persist beyond RTP, the specialised nature of these tests has limited their implementation in practice.

#### 1.3.5.1 Proposed underlying pathophysiology for motor changes following concussion

The increased injury risk observed post-RTP, appears linked to neuromuscular control deficits that linger following a concussion (4). Autonomic dysregulation is seen following concussion and has been shown to affect control of cerebral blood flow and cardiac rhythm (117). With regards to motor function, alterations in CNS function, more specifically, the primary motor cortex, may result in reduced metabolic activity, reduced cortical excitability, and/or an increase in intracortical inhibition. These changes are believed to alter muscle recruitment and affect movement patterns (6). Reductions in cortical excitability coupled with impaired intracortical inhibition has been associated with reductions in muscle tone and some reflex responses (76). This is likely responsible for delayed communication between the motor cortex and peripheral musculature (76). As a result, reductions in stiffness, particularly in response to perturbations, is seen amongst individuals who are deemed ready to RTP following concussion (6, 76). Reductions in lower-limb stiffness have been associated with non-contact lower extremity soft tissue and joint injuries, specifically when analysing landing mechanics (118). When analysing kinematic and kinetic characteristics of a single-leg jump post-concussion, quadriceps weakness and decreased time to peak vertical ground reaction force (PVGRF) was observed, implying reductions in lower limb stiffness (6). As a result, it appears these deficits may contribute to the increased risk of injury seen amongst returning athletes.

The motor cortex consists of the primary motor cortex, premotor cortex, and supplementary motor area (119). The neurons comprising the primary motor cortex share a direct connection with the motor neurons that send electrical signals to the skeletal muscles responsible for voluntary movement. The premotor cortex is responsible for pre-emptively activating the deeper postural muscles in preparation for movement, whereas the supplementary motor area assists with motor planning, coordination, and postural control (119).

Excitation and inhibition of neuronal activity is largely mediated by glutamate and gamma-aminobutyric acid (GABA), respectively (120). Glutamate is released from the pre-synaptic cell into receptors of the post-synaptic cell and increases electrical signals by allowing sodium and calcium (more positively charged) ions to enter the intercellular space (120). This causes depolarisation of the cell membrane, facilitating action potentials, and thus increasing excitability (120). Conversely GABA, which can be divided into $GABA_A$ and $GABA_B$, acts to inhibit excessive excitation by allowing (1) chloride (negatively charged ion) to enter the intracellular space ($GABA_A$) or (2) opening potassium channels to allow potassium to freely expel into the extracellular space ($GABA_B$) (120). Both mechanisms aim to restore the membrane resting potential by making it more negatively charged, which in turn inhibits further excitation (120).

Studies investigating motor cortex excitability following concussion have shown mixed results, with some finding no between group differences when assessed at 72-hours up to one-year post-concussion (121, 122). However, other studies have demonstrated reduced excitability and/or increased cortical silent period amongst recently concussed individuals up to 4-weeks post-concussion and amongst those suffering with chronic symptoms or with a history of multiple concussions (123-126). This may be observed through delayed motor responses, impaired coordination, and/or reductions in muscle force (123-126). Neurological differences have been seen up to 34-years post-concussion amongst healthy retired athletes (127). Changes in cortical excitability are believed to be associated with reductions in maximal skeletal muscle activation as well as increasing perceived sense of effort (6, 123, 128).

Alternatively, changes in intracortical inhibition may be a more realistic explanation for the changes in motor function seen post-concussion. Measures of intra-cortical inhibition are typically assessed by quantifying the cortical silent period (CSP) (120). This measure provides practitioners with the length of 'silencing' during voluntary skeletal muscle contractions (120). Studies have demonstrated increased CSP beyond one-year post-concussion (121, 122, 127, 129-131). Increased CSP duration has been associated with slowness of movement, in particular, increasing the period in which voluntary muscle contraction is supressed (121, 132).

It is hypothesised that such alterations in cortical function may impeded an athletes ability to plan, produce, or coordinate movements, thus leading to an increase in injury risk (6). However, it is important to note that concussion is a multifactorial injury, therefore it would be dismissive to only attribute the increase rate of injuries post-concussion solely to changes in motor control. Practitioners should also be aware of potential alterations in aspects such as decision making and planning of movement tasks, as these are also likely to contribute to increases in injury risk.

## 1.4 Current assessments used to determine return to play

Concussion management is driven by the consensus statement on concussion in sport (133), which is updated every four years, with the 6$^{th}$ iteration, published in June 2023 (134). Current management is highly dependent on the use of clinical assessments and self-reporting of symptoms (4, 70, 73). Current protocols for the assessment and management of concussion outlined in the consensus statement on concussion in sport promote the use of a multifaceted approach, assessing neurological, motor, and self-perceived symptoms (133). However, of concern is the questionable sensitivity of commonly used motor assessments beyond initial diagnosis (Table 1.2), with speculation surrounding the accuracy of appropriate RTP readiness (4) and the under-reporting of symptoms by players (22).

Table 1.2 Current side-line (SCAT6 and Child SCAT6) and office-based (SCOAT6) motor assessments used to assist with SRC diagnosis and/or RTP readiness

| Category | Assessment | Instrument/s | Outcome measure | Mechanism for error | Limitations |
|---|---|---|---|---|---|
| Coordination | Finger-to-nose | - | Ability to complete | Assesses cerebellar function<br><br>Coordinated movement involves interaction between temporal (timing) and spatial (joint position) awareness (135). Being able to adjust these aspects based on task requirements.<br><br>Deficits in cognitive-motor integration (CMI) results in reduced communication between brain networks (136) | Limited application to RTP requirements |
| Oculomotor | Balance Error Scoring System (BESS)<br>Modified BESS (mBESS) | Stopwatch<br>Foam pad (BESS) | Number of errors | Assesses static postural stability<br><br>Deficits in postural stability likely due to neurological (e.g. concussion) or lower limb impairment (e.g. ankle instability) (137, 138) | Subjective interpretation of errors<br><br>Number of errors influenced by fatigue, arousal, and/or learning effect, limiting its sensitivity immediately post-game-play or as an assessment integrated across the concussion management period (138)<br><br>Poor sensitivity beyond 1-week post-concussion |
| Gait | Timed tandem gait | Stopwatch | Time to complete<br>Number of errors | Hypoexcitability of cortical pathways or increased intra-cortical inhibition or primary motor cortex (139).<br>May impact ability to coordinate movements of increasing complexity (e.g. tandem gait). | Time to complete may not be sensitive enough to detect underlying motor abnormalities<br><br>Subjective interpretation of errors |
| Dual-task gait † | Dual-task timed tandem gait with addition of cognitive task (optional) | Stopwatch | Time to complete<br>Number of gait or cognitive errors | Reductions in executive function (139). Athletes may have reduced capacity to effectively divide attention between motor and cognitive tasks. | Time to complete may not be sensitive enough to detect underlying motor abnormalities<br><br>Subjective interpretation of errors |
| Aerobic capacity*† | Graded aerobic test of testing practitioners' choice | Equipment specific for chosen test | Score on selected test | Following concussion, athletes may experience exacerbation of symptoms due to exercise intolerance (140).<br><br>This may be due to abnormal autonomic nervous system functions and reductions in cerebral blood flow (141). | Optional assessment<br><br>Limited application of results if no aerobic capacity assessment performed prior to concussion |

† Highlights assessments that are optional for practitioners to complete; * Denotes aspects that are specific to the SCOAT6

### 1.4.1 Neurocognitive and symptom measures

Neurological and symptom assessments are commonly used during the acute stage and throughout the management of a concussion. Primary aims of these assessments are to establish timeframes when individuals report becoming asymptomatic and exhibit baseline neurocognitive scores. Approximately 38% of athletes may continue to experience neurological deficits beyond expected recovery timeframes (142). Common on-field assessments for concussion include; post-concussion symptom scale (PCSS), standardised assessment of concussion (SAC), immediate post-concussion assessment and cognitive testing (ImPACT), CogState, head injury assessment (HIA), King-Devick test, and the sports concussion assessment tool (SCAT) (1, 143). The primary aim of these assessments is to provide medical professionals with an informed clinical decision in regards to the removal of a player from a match or training situation following a head injury (108). However, of concern is the use of these assessments to inform RTP readiness, specifically considering their low sensitivity levels beyond the acute stages, lack of reliability, and learning effects (8, 144).

Extensive research into the efficacy of these assessments has been conducted, with results showing varied sensitivity and specificity values (143). Assessments such as ImPACT (sensitivity = 0.82-0.91, specificity = 0.69-0.89), SAC (sensitivity = 0.80-0.94, specificity = 0.76-0.91), PCSS (sensitivity = 0.48, specificity = 0.79), and the King-Devick test (sensitivity = 0.86, specificity = 0.90) have shown moderate to high sensitivity and specificity during the acute stages (1, 143). However, most studies indicate a return to baseline within 48 hours of sustaining a concussion (1). This quick resolution questions the sensitivity of these tests for detecting true recovery from concussion. Specifically, reliance on subjective interpretation, as well as the broad and somewhat inconsistent presentation of symptoms and cognitive impairments presents uncertainty as to how specific these assessments are in detecting underlying deficits (143). A systematic review investigated dual-task cost identified consistent reductions in cognitive accuracy amongst concussed participants up to 5 months post-concussion across included studies (107). Residual neurocognitive deficits may not be detected by traditional assessments, therefore falsely identifying some individuals as clinically recovered (107).

A common sideline assessment used in team sports is the SCAT (145). This assessment utilises a multifaceted approach, relying on self-reporting of symptoms, and subjective interpretation of observed balance errors and neurocognitive function by the assessor (146). Limitations surrounding the utility of this assessment are the lack of quantitative objectivity, inability to measure full recovery, and failure to account for mental health or learning disorders (7, 146). Due to the subjective nature of this assessment, test-retest reliability scores (ICCs) range from 0.49 to 0.63 (147). Additionally, this assessment has reported practice effects, therefore questioning its use in concussion management and in the RTP decision-making process (146). Limitations associated with these assessments including, poor reliability, the potential of learning effects, and the impact of fatigue and/or arousal on scores, may influence the short recovery timeframes observed.

### 1.4.2 Clinical assessments

Motor deficits are of primary concern during diagnosis and establishing RTP criteria for athletes. However a lack of objectivity and low sensitivity scores associated with these motor assessments (e.g. BESS, tandem gait) raises questions regarding their effectiveness for monitoring recovery following concussion (5). Typical motor assessments utilised for concussion diagnosis and management of RTP do not incorporate the use of specialised equipment, and therefore rely on subjective interpretation such as number of observed errors or time to complete the task (5). The Balance Error Scoring System (BESS) or a modified version, is the most widely used assessment to measure stationary balance (5). The modified version, typically integrated into sideline assessments, such as the SCAT, consists of three stances; double leg stance, single-leg stance on non-dominant leg, and tandem stance with non-dominant foot at the back (70). The BESS demonstrates high criterion validity amongst healthy individuals when compared to force platform sway (ICC = 0.78-0.96) (73). However, inter-rater reliability for total score has demonstrated only moderate reliability (ICC = 0.57 -0.74), potentially due to the binary scoring and subjective interpretation of results (73). The BESS has shown high specificity (0.91-0.97), however low sensitivity in detecting postural instability, particularly one-week post-concussion (0.34) (143). Outcomes of this assessment have been shown to be influenced by fatigue and group environments (143), as well as displaying a learning effect with repeated use (138). These assessments appear appropriate for the initial diagnosis of a concussion but are limited for the ongoing monitoring of symptoms.

The Rhomberg test (148), initially designed for older populations, aims at identifying static balance impairments with reduced visual input (70). Due to the design, no data exist for concussed individuals, creating uncertainty with respect to the utility of this assessment for this population.

### 1.4.3 Biomarkers

The use of biomarkers to measure the persistence of the metabolic disturbances associated with concussion and HIE is an emerging technique (143). Due to the large variance in symptom presentation amongst those with a concussion, reliance on symptom evaluation to guide RTP is limited. Biomarkers therefore may bridge this gap with the potential to be used as a prognostic tool (149). Specifically, the analysis of micro RNA biomarkers have the potential to identify signatures specific to the pathophysiological responses caused by concussion (149). Evidence has shown that analysis of blood, cerebrospinal fluid, and saliva may provide insight on injury severity (150). Biomarkers may be a more timely and accurate assessment tool for the diagnosis of concussion (143). This has been highlighted in previous studies, where salivary samples for suspected head injuries were successfully taken and processed during a common sideline assessment (149). A current limitation of these assessments is (1) their inability to distinguish between pathophysiological processes of concussion, and those of other neurological and/or mental health disorders, and (2) possibility that differences in individual genetics influence the response of markers (143). Additionally, biomarkers have only currently been used as diagnostic tools (143), therefore questioning their consistency, as well as ability to be used for management as well as detecting any motor or cognitive impairments (67). Although these techniques are only emerging in the literature, the use of salivary and blood-based biomarkers have shown the potential to aid in the diagnosis of concussion (149). Despite this, the cost and need for specialised equipment may limit their widespread use in diagnosing and managing concussion in an applied sports setting.

### 1.4.4 Laboratory assessments

Laboratory assessments appear superior at detecting ongoing concussion symptoms, when compared to subjective tests. These assessments have largely focused on static and dynamic balance, gait characteristics, and dual-task performance (7, 75, 107, 110, 113, 114, 151). Balance assessments utilising force platforms and centre of pressure metrics have proven to be effective in falls risk populations such as stroke patients, older adults, and those with Parkinson's disease (152, 153). Specifically, the sensory organisation test makes use of a force platform and harness cage. The aim of this assessment is to disrupt sensory information through manipulation of spatial awareness (70). Individuals are required to maintain quiet stance during each trial, with centre of pressure metrics used to calculate postural sway (70). This assessment has demonstrated higher sensitivity (0.62) post-concussion when compared to the BESS, primarily due to its objective nature (99). However, poor test-retest reliability, minimal research into concussed populations and the need for specific equipment, questions the use of this assessment following SRC (70). Extending from the Sensory Organisation Test (SOT), studies have implemented the use of force platforms to assess results from the BESS. Specifically, COP sway velocity in the medio-lateral axis and COP sway velocity in the anterior-posterior axis have been shown to be sensitive in detecting concussion during double-leg and tandem stance respectively (146). Results support the use of these metrics derived from force platforms, highlighting their sensitivity to detect postural deficits up to 6 months post-concussion (146).

Assessments utilising force plates, pressure sensitive walkways, and motion analysis have shown good test-retest reliability for healthy adolescents through to older adults (154, 155). Gait changes have been assessed through centre of mass (COM) displacement, peak velocity in both the anterior-posterior and medio-lateral directions, gait velocity, stride length, cadence, and step width (107, 114, 143). Tandem gait assessments claim to be clinically viable assessments with complex motor control requirements (114). Previous studies have shown deficits in tandem gait amongst those with concussion, despite similar modified BESS scores (114). Concussed paediatric and collegiate athletes have demonstrated slower tandem gait when compared with healthy controls (114, 156). Tandem gait assessment has been reported to have a sensitivity of 0.63 and specificity of 0.65 for detecting concussed collegiate athletes

(156). Preliminary evidence suggests tandem gait analysis is a clinically viable assessments for practitioners to run and may offer better diagnostic value when compared to traditional quiet stance assessments (e.g. BESS)  (114). It is proposed that the task characteristics result in a larger emphasis on the ankle control strategy; that is the active and passive mechanisms of the ankle musculature to maintain balance during quiet stance (157, 158). Therefore, individuals with underlying neuromuscular deficits may present with an impaired capacity to stabilise using the ankle strategy during tandem stance, and potentially place more reliance on the hip control strategy, resulting in more hip and trunk movement to maintain balance (146, 157, 158).

Although clinically viable, the lack of sensitivity of single-task gait assessments has led to the integration of dual-task assessments. As previously mentioned, pathophysiological processes following concussion may result in impaired executive function and therefore may affect one's ability to divide attention between concurrent tasks (110, 113-115). Under dual-task conditions, a greater number of cognitive errors were reported for those returning from concussion, with their slower walking velocity also linked to increased medio-lateral COM sway (75, 114). The sensitivity and specificity of single-task and dual-task tandem gait along a linear 3-metre course in concussed adolescents has been examined (114). Results indicated that both single-task and dual-task scenarios showed moderate to high sensitivity and specificity scores. Specifically, single-task gait with a cut-off time of sixteen seconds had a sensitivity of 0.87 and specificity of 0.72. Moreover, dual-task gait with a cut off time of twenty-two seconds had a sensitivity of 0.84, and specificity of 0.72 (114). These results provided evidence for the support of single-task and dual-task gait assessments to identify concussion ten days post injury. The use of more functional athletic tasks is limited, but one study has shown longer time to stabilisation on a single leg hop test in concussed athletes compared with controls (159). Collectively these results suggest that more complex, dynamic motor tasks may provide greater utility at detecting concussion compared to static assessments.

Laboratory assessments are ideal in research settings and provide practitioners with useful information into the aetiology of concussion. However, they have low ecological validity for assessment of field-based sports, due to the controlled environmental conditions in the laboratory required that do not replicate the field-based sports performance (160). Additionally, equipment requirements and facilities used are costly and usually not feasible for team sports to access. In concussion research, equipment used such as force plates, pressure sensitive walkways, and video motion analysis have shown high reliability and sensitivity, however barriers such as those mentioned above, have hindered their use amongst team sport athletes in an applied sport setting (160). The limitations noted therefore compromise the utility of laboratory assessments for rehabilitating concussed athletes.

### 1.4.5 Inertial measurement units (IMU) instrumented assessments

Inertial measurement units (IMU) are portable devices that consist of a triaxial accelerometer, gyroscope, and magnetometer that allow the measurement of acceleration, angular velocity, and magnetic field strength (to orient data with the Earth's axes), respectively (161). Collectively, these devices allow for the detection of gait events and spatiotemporal parameters such as contact time, flight time, stride and step frequencies, and stride and step lengths (162, 163). Sampling frequencies of these devices may reach 1000 Hz, allowing for the detection of discrete events that occur during motor tasks (162). However, lower sampling frequencies of 100 Hz are commonly integrated into the wearable sensors worn by team sport athletes during training and competition (162).

As mentioned, triaxial accelerometers measure linear accelerations in three directions; anterior-posterior, medio-lateral, and vertical axes (164). Combined with knowledge of the athlete's mass, accelerometers can also provide the user with insight into the forces experienced by an individual during movement tasks. Various commercially available wearable microtechnology devices provide accelerometer loading metrics (e.g. PlayerLoad™) (165). These metrics aim to capture resultant acceleration and provide a vector magnitude of changes in acceleration across all three axes (165-167), typically presented as an arbitrary unit (167). In addition, instantaneous and accumulated accelerations within each plane can also be obtained, providing greater insight into how accelerations are accumulated (167). With increased medio-lateral movement observed in concussed athletes (108) across various tasks, quantifying accelerations from IMUs along the individual axes could be beneficial for detecting and monitoring symptoms.

Previously, IMUs have been used to monitor motor performance following concussion (144, 145, 168, 169). An IMU-instrumented Y balance test amongst elite rugby players with IMUs placed on the lumbar spine, showed changes in dynamic balance with reductions in reach distance in the posterior-medial and posterior-lateral directions in concussed players (144). Other balance assessments using head- (168) and lumbar-mounted (145) accelerometry have shown increases in sway volume following concussion. In more dynamic tasks, like dual-task gait walking assessments, lumbar-mounted IMUs have shown reductions in peak accelerations and angular velocities up to 2-months post-concussion (169). This was evident during heel strike and early single-support phase of walking gait (169). Collectively, it appears that IMUs may be able to detect changes in balance and gait during walking following concussion and offer a portable, more cost-effective method of monitoring symptoms compared to laboratory or traditional clinical assessments. Notably though, no assessments have been performed at velocities above walking pace, which would seem important given the risk of subsequent injury post-concussion for athlete populations. Post-concussion movement changes may be exacerbated by high movement velocities.

Wearable microtechnology has long been used in team sports as a means of monitoring athlete external loads (170). These devices, worn in padded vests on the thoracic spine, typically house a global navigational satellite system (GNSS) chip and IMUs comprising a triaxial accelerometer, gyroscope and magnetometer. The IMUs within these devices have shown potential in monitoring neuromuscular fatigue (171). Previous research analysing the use of accelerometry to measure running load index (RLI), a measure of the ratio between velocity load and force load (172), identified strong relationships with leg stiffness, evident through changes in RLI vertical measures. Additionally, RLI had high sensitivity to neuromuscular fatigue monitored via a counter movement jump (CMJ) (173). Reductions in vertical accelerations and differences in accumulation of load per minute have also been observed in professional Australian rules football players with neuromuscular fatigue (171). Small to moderate reductions in all axes, with specific emphasis on vertical accelerations have also been observed during set-paced 50 m running efforts in Australian football players (174). A similar study conducted on professional A-League soccer players identified changes in movement strategies in small-sided games amongst fatigued players (175). Specifically, an increased contribution of medio-lateral PlayerLoad™ was identified (175). Others have shown accelerometer load reductions across repeated sprints as players become fatigued (176) and that accelerometer axes offer appropriate signal-to-noise ratios for detecting changes in movement strategies when fatigued (177). These results provide support for the use of triaxial accelerometers housed within microtechnology devices, to successfully measure changes in movement strategies amongst fatigued and non-fatigued players. It is worth noting that there are potential issues with thoracic worn devices. If the unit is mounted within a vest or jersey rather than affixed to the skin and the unit is located further away from the ground, this potentially introduces signal noise and may diminish changes in acceleration profiles (162, 167).

Of particular interest is the difference in validity and reliability between varied placements of IMU devices for monitoring human gait during movement tasks. Historically, there has been hesitation in using the upper body to measure spatiotemporal gait metrics, due to accelerometers being designed to measure the limb they are attached to (162). However, a recent meta-analysis assessing reliability and validity of different placements concluded that contact time and step frequency derived from IMUs placed on lumbar spine, tibia, and foot did not significantly differ from criterion (162). It was additionally noted that placements outside of these sites may still be valid however the need to classify stride events is necessary (162). In regards to the reliability of different IMUs and different placement sites, a systematic review identified mixed results (178). Intra-session test-retest reliability was dependent on device location. Foot-mounted IMUs demonstrated *good* to *excellent* reliability for measures of contact time, flight time, step frequency, and step length (162). Lumbar-mounted IMUs demonstrated *good* to *excellent* reliability for measures of flight time, step frequency, vertical GRF, and vertical stiffness. Measures of contact time were questionable, ranging from *poor* to *excellent* reliability (162). Thoracic-mounted IMUs have also shown *good* to *excellent* inter-day reliability for running gait of increasing velocities (3-8 $m \cdot s^{-1}$) (179).

Previous studies assessing validity and reliability of thoracic-mounted IMUs to assess physical activity has been reported. This device location has shown to be a valid measure of step time, step frequency, step length, and vertical stiffness over increasing velocities (3-8 $m \cdot s^{-1}$) (179). In regards to sport-specific applications, a 100 Hz accelerometer demonstrated high between-device reliability (CV 1.9%) (164), and minimal noise (CV <2%) for measurement of physical activity in team sports. Another study identified *small* to *moderate* (0.47 – 0.68) standardised typical error when using IMUs to monitor fatigue in rugby union players (173). Therefore highlighting utility of these devices to detect discrete differences in physical activity metrics (164). Comparison between scapula and COM (lumbar) derived PlayerLoad™ metrics has also been investigated (167). Both locations displayed moderate to high test-retest reliability (ICC = 0.80 – 0.97, CV 4.2 – 14.8%) (167). Player load derived from COM-mounted IMU was significantly higher when compared with measures taken from the scapula (223.4 ± 42.6 vs 185.5 ± 26.3, p = 0.001). Analysis of individual contributions showed the COM measures were higher for medio-lateral Player Load (COM 26.5% ± 4.9%, p = 0.001, scapula 20.4% ± 3.8%), but lower for vertical PlayerLoad™ contributions (COM 49.5% ± 6.9%; P = .001, scapula 55.7% ± 5.3%) (167). Although both placements showed moderate to high test-retest reliability, and convergent validity when compared with exercise intensity, it was highlighted that caution be taken when measuring PlayerLoad™ from scapula positioned IMUs (167).

Although metrics derived from the scapula and/or thoracic region (via manufacturer bib) have their limitations, it would be dismissive to ignore the potential benefits of using this device placement to aid in concussion management. Given the data is often already being collected in teams, it may prove to be the most efficacious option to collect gait data from athletes. This data may then be used to assess movement profiles of athletes following injury.

## 1.5 Validity and reliability

The terms validity and reliability are concepts used throughout this thesis to understand the utility of microtechnology for concussion management. In general terms, validity refers to an ability to measure what is intended to be measured (180). More specifically, there are three types of validity: logical, criterion, and construct. This thesis will primarily focus on construct validity; that is the degree in which a protocol measures a construct. Construct validity will identify how well the selected assessments and tools differ between concussed and healthy populations. Informing the extent in which concussed athletes demonstrate movement deficits post-concussion and the time-course of these deficits. Sensitivity refers to a protocol or measure being able to correctly identify those with a condition (i.e., concussion), while specificity refers to an ability to correctly identify those without a condition (180, 181). Reliability refers to the consistency of a measure and its repeatability over time (181). Notably, the assessment of both absolute reliability, that is the agreement of repeated measures, and relative reliability reflecting within-subject variance over time (182), are important variables to consider when assessing the reliability of IMUs to measure gait parameters during walk, jog, and run tasks on grassed surfaces.

## 1.6 Statement of the problem and general aims

Previous literature has highlighted potential complications following premature RTP from concussion in team sport athletes. Of particular concern is the increased injury incidence amongst those returning from concussion. Previous analysis of clinical and laboratory assessments, as well as emerging measures such as biomarkers and microtechnology has highlighted both their utility and limitations as primary tools for the assessment and management of concussion in team sport athletes. Of interest within this thesis was the use of thoracic worn IMUs and their potential to allow continuous measurement of movement profile recovery in recently concussed players. This thesis aimed to investigate the utility of ankle, lumbar, and thoracic IMUs through assessing their reliability and ability to discriminate between concussed and non-concussed players across a range of movement velocities.

## 1.7 Aims of the current research

The aims of this thesis were to (1) Consolidate the reliability and validity of motor function assessments currently used in concussion management and further summarise their feasibility for clinical use. (2) Assess the reliability of ankle, lumbar, and thoracic mounted devices to measure gait parameters on grassed surfaces. (3) Determine if lumbar- and thoracic-mounted IMUs can detect changes in movement strategies amongst athletes returning to play following a diagnosed sports-related concussion (SRC). Findings within this thesis provide insight into alternative solutions to monitor motor strategy changes following SRC in a manner that is practical for clinicians and practitioners. Specifically, this thesis will:

1. Systematically consolidate the reliability and validity of motor assessments currently used within concussion management and further summarise their feasibility for practitioners (Study 1; Systematic Review).
2. Investigate the test-retest reliability of ankle-, lumbar-, and thoracic-mounted IMUs for measuring discrete stride variables on grassed surfaces (Study 2).
3. Determine whether movement differences exist between concussed and healthy athletes during gait tasks when derived from lumbar- and thoracic-mounted IMUs (Study 3). This will include determining the:
    a. Time course of movement differences; and
    b. Influence of movement speed or pace condition

Upon completion of the studies included within this thesis, this thesis will provide insight into the use of wearable technology during gait tasks performed on natural grassed surfaces to provide clinicians and practitioners with objective and cost-effective ways to monitor recovery post-concussion. The schematic overview of the studies included within this thesis is presented in Figure 1.1

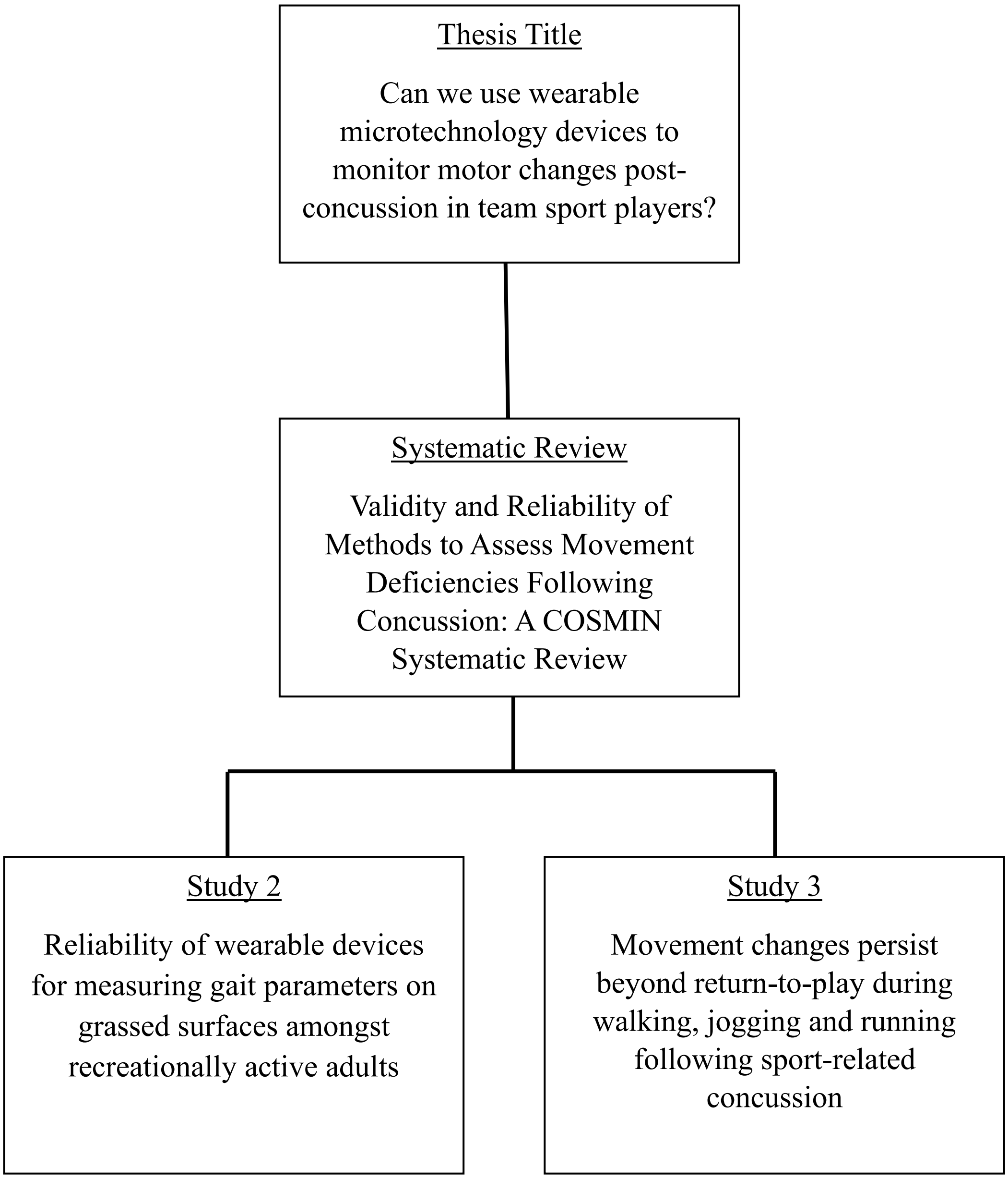


Figure 1.1 Schematic overview of the studies that comprised the thesis.

# Chapter 2

# Systematic Review

**Validity and reliability of methods to assess movement deficiencies following concussion: A COSMIN systematic review**

This thesis aimed to determine the reliability and sensitivity of wearable devices to monitor movement strategy changes following SRC. Notably, can wearable devices be used to complement current concussion management protocols, providing a more comprehensive assessment of readiness for RTP. Given the aims of this thesis, it was first necessary to understand what evidence was available regarding the reliability and/or validity of the different motor assessments currently available to clinicians and/or practitioners to diagnose and manage RTP following concussion.

This systematic review has been published in Sports Medicine Open. Full reference details are:

Dunne, L.A.M., Cole, M.H., Cormack, S.J. *et al.* Validity and Reliability of Methods to Assess Movement Deficiencies Following Concussion: A COSMIN Systematic Review. *Sports Med - Open* **9**, 76 (2023). https://doi.org/10.1186/s40798-023-00625-0

Please refer to Appendix 1 for evidence of publication.

**Abstract**

*Background:* There is an increased risk of subsequent concussion and musculoskeletal injury upon return to play following a sports-related concussion. Whilst there are numerous assessments available for clinicians for diagnosis and during RTP following concussion, many may lack the ability to detect these subclinical changes in function. Currently, there is no consensus or collated sources on the reliability, validity and feasibility of these assessments, which makes it difficult for clinicians and practitioners to select the most appropriate assessment for their needs. *Aim:* This systematic review aims to (1) consolidate the reliability and validity of motor function assessments across the time course of concussion management, and (2) summarise their feasibility for clinicians and other end-users. *Methods:* A systematic search of five databases was conducted. Eligible studies were (1) original research; (2) full text English language; (3) peer-reviewed with level III evidence or higher; and (4) assessed the validity and/or test-retest reliability of lower limb motor assessments used to diagnose or determine readiness for athletes or military personnel who had sustained a concussion to return to play or duty. Acceptable lower limb motor assessments were dichotomised into instrumented and non-instrumented and then classified into static (stable around a fixed point), dynamic (movement around a fixed point), gait, and other categories. Each study was assessed using the COSMIN checklist to establish methodological and measurement quality. *Results:* A total of 1270 records were identified, with 637 duplicates removed. Titles and abstracts of 633 records were analysed, with 158 being retained for full text review. A total of 67 records were included in this review; 37 records assessed reliability, and 35 records assessed validity of lower limb motor assessments. There were 42 different assessments included in the review, with 43% being non-instrumented, subjective assessments. Consistent evidence supported the use of instrumented assessments over non-instrumented, with gait-based assessments demonstrating sufficient reliability and validity compared to static or dynamic assessments. *Conclusion:* These findings suggest that instrumented, gait-based assessments should be prioritised over static or dynamic balance assessments. The use of laboratory equipment (i.e., 3D motion capture, pressure sensitive walkways) on average exhibited sufficient reliability and validity; yet demonstrate poor feasibility. Further high-quality studies evaluating the reliability and validity of more readily available devices (i.e., inertial measurement units) are needed to fill the gap in current concussion management protocols. Practitioners can use this resource to understand the accuracy and precision of the assessments they have at their disposal to make

informed decisions regarding the management of concussion. *Registration:* This systematic review was registered on PROSPERO (reg no. CRD42021256298).

**Keywords:** sports-related concussion, motor, assessment, validity, reliability

**Key Points**

Commonly used subjective static assessments such as the Balance Error Scoring System (BESS) displayed insufficient test-retest reliability and construct validity for the detection of sports-related concussion (SRC).

Instrumented static balance assessments using laboratory equipment (i.e., Force plate) or portable microtechnology (i.e., Inertial measurement units) demonstrated better test-retest reliability and construct validity compared to subjective assessments. However, all static balance assessments displayed a poor ability to detect persistent symptoms of SRC beyond acute stages (> 2-weeks post).

Instrumented dynamic assessments demonstrated sufficient test-retest reliability. The instrumented Y-balance test demonstrated sufficient sensitivity on adult populations, however poor specificity.

Instrumented and non-instrumented gait assessments displayed sufficient test-retest reliability and construct validity. The addition of a cognitive task (dual-task) improved sensitivity.

Laboratory assessments display sufficient reliability and validity, however poor ecological validity for the assessment of field-based sports due to the controlled environmental conditions. Associated costs, equipment, and personnel also limit the utility of these assessments for team sport athletes.

Clinicians are encouraged to implement instrumented or non-instrumented dynamic balance or gait assessments based on the individual needs and abilities within their setting.

If practitioners do not have the resources to perform instrumented tests, it is recommended that they consider the reliability and validity issues that potentially limit the simpler test options, with gait assessments recommended over static or dynamic

## 2.1 Background

Concussions, otherwise referred to as mild traumatic brain injury (mTBI), is described as a transient disturbance of brain function (1) and are common injuries in contact sports, such as rugby league (183), and in certain occupations, such as military personnel (184). Concussions are caused by transfer of energy across the brain due to direct (collision) or indirect (whiplash mechanism) trauma to the head and/or neck (2, 45). Such impacts cause disruptions in normal cellular function, resulting in an 'energy crisis' (2, 45, 48, 57, 122, 185), with symptoms typically including headache, nausea, poor coordination, vision deficits, and behavioural abnormalities such as irritability or depressive mood states (2, 47). Given the multiple symptoms that present following a concussion, monitoring recovery can be complex for clinicians and practitioners.

To account for the multitude of symptoms experienced, a variety of assessment tools are made available for clinicians (7). Across numerous sports, athletes diagnosed with a concussion are guided through a graduated return to play (RTP) process by a medical practitioner and/or rehabilitation staff. Progress through the staged RTP is primarily based upon symptom resolution at rest and during exertion as well as a return to pre-concussion baseline for cognitive and motor scores (4, 70, 73, 186). Of concern, however, is the ambiguity surrounding diagnostic tools and more specifically, the lack of evidence supporting their implementation in the latter stages of concussion management. For example, the common subjective balance assessments used by clinicians (e.g., BESS and tandem gait) (187) may lack the resolution to detect changes in function that can linger post-concussion. There appears to be an increased risk of subsequent concussion and musculoskeletal injuries, up to 12 months following SRC (102, 112, 188), which may be linked to lingering motor deficits (99); indeed suggesting subclinical changes remain beyond RTP clearance that are poorly detected by many of the assessments readily available to clinicians (100, 112, 188). Reliance on diagnostic tools as a means to evaluate recovery in conjunction with the subjective nature of many clinical assessments may explain why subtle, underlying motor changes go largely undetected (189). Due to this concern, it is important to understand how post-concussion changes in motor performance can be monitored more effectively, thus allowing clinicians to make decisions based on sound objective data as well as clinical judgement.

To minimise the risk of incorrect recovery diagnosis, assessments need to demonstrate clinically acceptable reliability and validity, whilst also being feasible to conduct. Reliability refers to an instrument's ability to produce consistent measures across multiple time points, thus ensuring change in score is attributed to changes in performance as opposed to instrument errors (181, 190). Validity can be broken into three categories; logical, criterion, and construct (180). For this review, only construct validity has been reported. That is an instrument's ability to correctly diagnose concussed and non-concussed populations. The higher the sensitivity and specificity of an instrument, the better its ability to classify those with and those without concussion (180). Feasibility is also vital to consider when selecting a test, the time, necessary resources and expertise required as these will influence which tests can be administered.

Numerous lower limb motor assessments are reported in the literature to monitor impairments following concussion, with varying time, expertise and equipment requirements. Despite this, there is no consensus or collated sources on the reliability, validity and feasibility of these assessments, which makes it difficult for clinicians and practitioners to select the most appropriate assessment based on needs and time since concussion. This systematic review aims to (1) consolidate the reliability and validity of motor function assessments across the time course of concussion management, and (2) summarise their feasibility for clinicians and other end-users. The purpose is to provide clinicians with evidence to support the utility and practicality of selected assessments and identify potential gaps in the current management of concussion.

## 2.2 Methods

### 2.2.1 Search strategy

This systematic review was structured in accordance with the Preferred Reporting Items for Systematic Reviews and Meta Analyses (PRISMA) statement (191) and registered on PROSPERO (reg no. CRD42021256298). Five academic databases, including SPORTDiscus, CINAHL, Web of Science, Medline, and Scopus were systematically searched from earliest record to May 17, 2023. Eligible studies were identified through searching titles, abstracts, and keywords for pre-determined search terms (Table 2.1). References were extracted from each database and imported into a reference manager (EndNote X20, Clarivate Analytics, London, United Kingdom) before removing any duplicate articles. Subsequently, to allow simultaneous, blinded screening, articles were imported into Covidence (www.covidence.org; Melbourne, Australia), an online tool for systematic reviews. Titles and abstracts were analysed by one reviewer (LD); the full texts of remaining studies were then assessed by two reviewers (LD and RJ). Where any conflicts arose, the two reviewers met to determine study eligibility.

Table 2.1 Search terms used for review; search 1 to 5 was combined with the operator 'AND', search 6 was combined with the operator 'NOT'

| | |
|---|---|
| **Search 1** | sport OR athlete OR player* OR military OR soldier OR "service men" OR "service member*" |
| **Search 2** | concuss* OR "sports related concussion" OR "sports-related concussion" OR mTBI OR "mild traumatic brain injury" OR "sport-induced concussion" OR "sport induced concussion" OR "mild head injury" |
| **Search 3** | assessment* OR test* OR evaluat* OR analysis OR examination OR outcome OR measure |
| **Search 4** | COP OR centre of pressure OR center of pressure OR gait OR movement OR single-task OR single-task OR stiffness OR motor OR neuromuscular OR IMU OR "inertial measurement unit" OR accelerom* OR landing OR dynamic balance |
| **Search 5** | validity OR reliability OR sensitivity OR specificity OR "test-retest reliability" |
| **Search 6** | "motor accident" OR "car accident" OR "car crash" OR "wreck" OR "vehicle accident" OR "vehicle crash" OR "truck crash" |

### 2.2.2 Eligibility criteria

Eligible studies must have (1) been original research articles; (2) been full text articles written in the English language; (3) been peer-reviewed articles with level of evidence equal to or greater than level III (192); (4) assessed the validity of lower limb motor assessments used to diagnose or determine RTP readiness for athletes or military personnel who had sustained a concussion or; (5) assessed the test-retest reliability of lower limb motor assessments used for concussion management amongst healthy athletes. Acceptable lower limb motor assessments were classified into four categories: static, dynamic, gait, and other. Static balance assessments included tasks in which individuals remained in a fixed point during various stances (e.g., BESS) where postural sway or number of balance errors were the outcome variables. Dynamic balance assessments included any task that required movement (e.g., limb excursion) from an individual, while remaining at a fixed point (e.g., Y-balance test). Gait assessments comprised of any task that required locomotion with both temporal and/or spatial parameters measured. Assessments that were specific for sport or military tasks were categorised as other. Further categorisation was performed with assessments being classified as non-instrumented (subjective scoring or use of basic equipment [i.e., Stopwatch]) or instrumented (objective [i.e., accelerometers]).

For studies to be included as reliability studies, they must have assessed the test-retest (intra-class correlation coefficient (ICC)) or inter-rater reliability of an assessment using healthy athletes. For validity, studies must have assessed the between group differences of a lower-extremity motor task in a case-control study or shown the predictive performance of the measure to diagnose concussed and healthy participants (i.e., area under the curve (AUC), sensitivity, specificity). Reference lists from eligible studies were manually examined for any studies missed during initial search. Selected studies were then screened and assessed for eligibility. Commentaries, letters, editorials, conference proceedings, case reports, conference abstract, or non-peer reviewed articles were excluded. Studies examining animal or biomechanical models of brain injury, were also excluded from analysis.

### 2.2.3 Data extraction and analysis

Data from eligible studies were extracted into Covidence. Data pertaining to study characteristics and protocols were first extracted from eligible studies. All relevant outcome measures (reliability and/or validity measures) were extracted from each study. Data were categorised according to: assessment type (e.g., static, dynamic, gait), and relevant findings being reliability and/or validity (e.g., sensitivity, specificity). Due to the heterogeneous nature of the findings a meta-analysis was not performed.

### 2.2.4 Quality assessment

To assess the methodological quality and the clinical reported outcome measurements (ClinROMs; reliability and validity), the Consensus-based Standards for the selection of health Measurement Instruments (COSMIN) Risk of Bias tool for outcome measurement instruments (193) and the COSMIN guideline on Risk of Bias to assess quality of studies on reliability and measurement error, that is the variability between repeated measures, was used (194). The COSMIN checklists were developed to quantitatively assess the methodological quality of studies and the ClinROMs evaluated. The first step involved rating the methodological quality for each study, which was assessed against nine measurement properties: content validity, internal structure (structural validity, internal consistency, and cross-cultural validity), reliability, measurement error, criterion validity, hypotheses testing for construct validity, and responsiveness. Each measurement property was assessed using a four-point grading scale; v*ery good*, where the model or formula was described and matched the study design; *adequate*, where the model or formula was not described, or did not match the study design; *doubtful*, where no evidence of systematic difference was provided; and *inadequate*, where calculation was deemed not optimal. Overall methodological reporting quality was determined using the 'worst score counts' approach (193, 195). Feasibility of the assessment tool is no longer included within COSMIN's measurement properties as it does not refer to the quality of an outcome measurement instrument. We highlighted the feasibility of an instrument, by reporting the interpretability of the outcome, time to complete, and equipment and expertise required. The second step was to rate the ClinROMs from each study (validity and/or reliability values) using the COSMIN criteria for good measurement properties guideline (195). A rating of *sufficient* (+), *insufficient* (-), or *indeterminate* (?) was given for each assessments measurement property based on the statistical outcome measures for each measurement property (194). Two authors (LD and RJ) independently assessed the methodological quality and measurement property of all studies; any disagreements were discussed by these authors.

## 2.3 Results

### 2.3.1 Search results

The systematic search retrieved 1270 results from five academic databases, of which 637 duplicates were removed. Titles and abstracts of remaining 633 studies were screened, with 475 not meeting eligibility criteria. Full text review was conducted on the remaining 158 studies, with 112 deemed ineligible. A total of 46 studies were eligible, with an additional 21 included via the manual screening of reference lists. Therefore, this review included a total of 67 studies. The identification process is outlined in Figure 2.1.

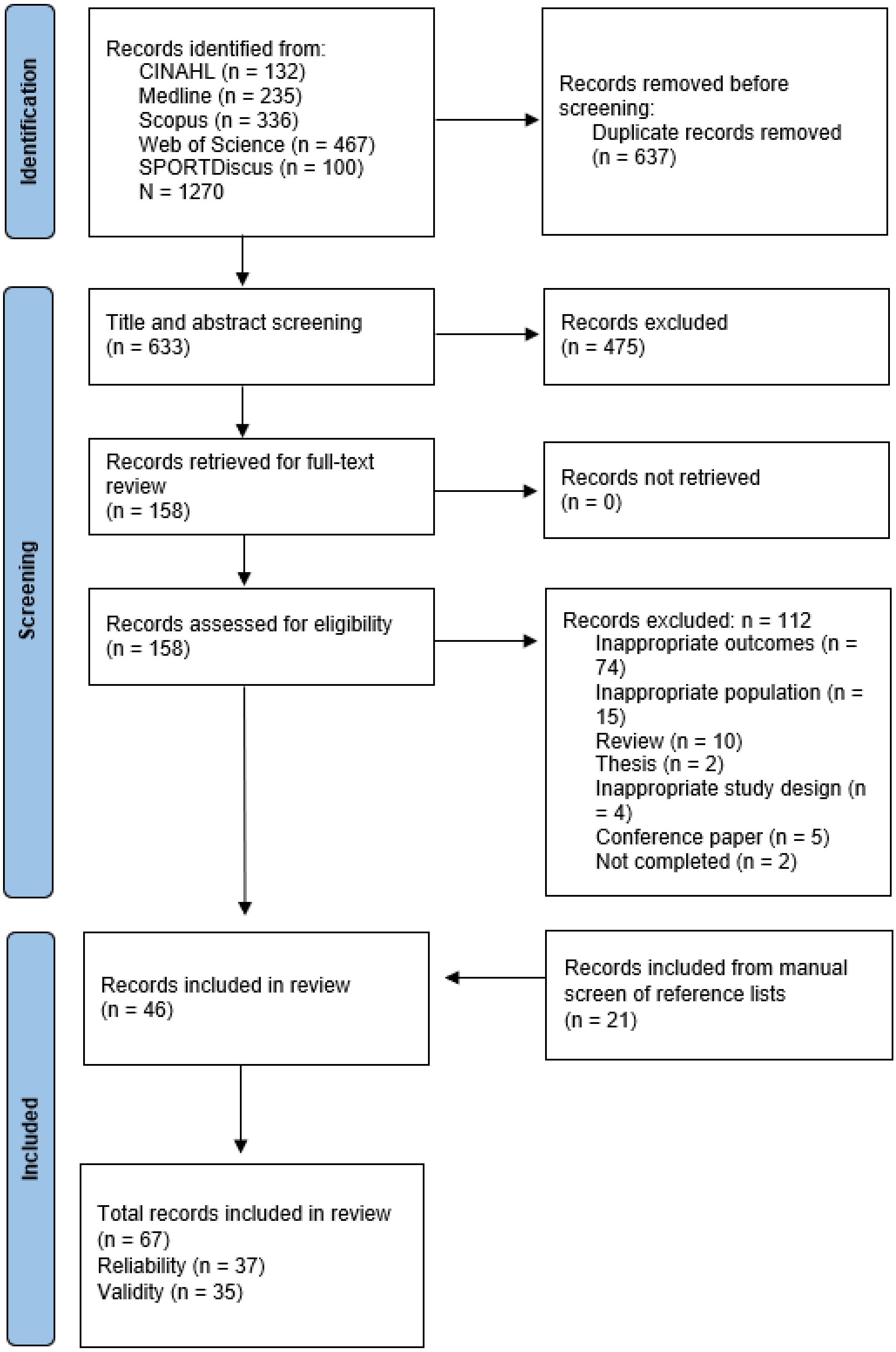


Figure 2.1 PRISMA flow-chart depicting steps taken in the search strategy

### 2.3.2 Research quality

The quality of research investigating the reliability and/or validity of lower limb motor assessments for concussion management was variable, with methodological reporting quality ranging from *inadequate* to *very good*. Measurement property quality for all studies ranged from *sufficient* to *indeterminate* (See Tables. 2.6 – 2.20).

### 2.3.3 Study characteristics

#### 2.3.3.1 Reliability

Studies were conducted on healthy adults (n = 29) and minors (n = 8), with a total sample size of 6,888. The most common assessments were the BESS and tandem gait (instrumented and non-instrumented), each representing 15% of all assessments. A summary of study characteristics is presented in Table 2.2, 2.3, 2.4, and 2.5. Full table of study characteristics are presented in Tables 2.6 through to Table 2.12.

There were 22 different lower limb motor assessments used across 37 different studies; 12 studies assessed the reliability of more than one test, one study assessed reliability for adults and minors (See Supplementary Table 1). Assessments were categorised as static balance (n = 20 studies, 9 different assessments), dynamic balance (n = 5 studies, 4 different assessments), gait (n = 13 studies, 9 different assessments), or other (n = 1 study, 1 assessment). Studies were further sub-divided based on type of reliability; test-retest (n = 34 studies, 20 different assessments) or inter-rater (n = 5 studies, 5 different assessments) and instrumented (n = 13 assessments) or non-instrumented (n = 9 assessments).

#### 2.3.3.2 Static balance assessments

For static balance assessments, test-retest correlations ranged from 0.13 to 0.94 with measurement property quality ranging from *doubtful* to *adequate*. Outcome variables for non-instrumented assessments included time and number of errors. Instrumented assessments reported number of errors, center of mass (COM) displacement, and center of pressure (COP) displacement. Time between assessments ranged from same day to 20-months, with a tendency for poorer reliability over longer periods. Assessments included BESS (n = 5), instrumented BESS (n = 2), modified BESS (mBESS) [double leg, single leg, and tandem stance on firm ground] (n = 3), instrumented mBESS (n = 1), single leg stance (n = 2), instrumented single leg stance (n = 1), double leg balance using accelerometers (balance accelerometry measure (BAM)) (n = 2), double leg balance on a portable force plate (balance tracking system) (n = 1), double- and single-leg balance (SWAY balance mobile application) (n = 1), and the Sensory Organization Test (SOT) (n = 2). The BESS demonstrated *sufficient* reliability when conducted with one trial (ICC = 0.60-0.78). However, reliability was improved when double leg stance was removed and two to seven trials were performed (ICC = 0.83-0.94). Instrumented BESS using a force plate and Wii Balance Board (0.88-0.89), and the balance tracking system (ICC = 0.92) also displayed *sufficient* reliability over seven and 15-day periods, respectively (196, 197). The BESS and mBESS showed improved reliability with increased number of trials (198). It is imperative to note that, while studies report improved reliability with increased number of trials, these assessments are routinely performed only once in clinical practice. In summary, a minimum of 2-trials on 4 conditions (excluding double leg variations) of the BESS displayed *sufficient* test-retest reliability over a 7-day period (199). The balance tracking system utilising a force plate also displayed *sufficient* reliability in addition to offering clinicians more in-depth, objective analysis (196).

2.3.3.3 Dynamic balance assessments

For dynamic balance assessments, test-retest correlations ranged from 0.32 to 0.99, with measurement property quality ranging from *doubtful* to *adequate*. Outcome variables included time, number of errors, COM displacement, and COP displacement. Time between assessments ranged from same day to 11-months, with a median of seven days; with a tendency for poorer reliability over periods greater than 10-days. Assessments included instrumented Y-balance test (n = 1), clinical reaction time (n = 1), instrumented limits of stability test (n = 2), and the dynamic postural stability index (DPSI) (n = 1). The most reliable assessments were the instrumented Y-balance test (ICC = 0.76 to 0.99), which performed same-day test-retest reliability (200) and the instrumented limits of stability test (ICC = 0.95 to 0.96), with tests conducted 7-days apart (201). Both assessments provided clinicians with consistent objective measures across trials.

#### 2.3.3.4 Gait assessments

For gait assessments, test-retest correlations ranged from 0.10 to 0.99, with measurement property quality ranging from *doubtful* to *adequate*. Outcome variables for non-instrumented assessments included time or number of errors. Instrumented assessments reported, COM displacement, COP displacement, and spatiotemporal metrics. Time between assessments ranged from same day to 11-months, with a median of seven days and a tendency for poorer reliability over periods greater than two weeks. Assessments included tandem gait (n = 6), instrumented gait (n = 7), instrumented dual-task gait (n = 2) dual-task tandem gait (n = 2), instrumented dual-task tandem gait (n = 2), timed up and go (TUG) (n = 1), and walking on a balance beam (n = 1). Most gait assessments displayed *sufficient* test-retest reliability; non-instrumented assessments displayed *insufficient* reliability across periods extending greater than two months. Instrumented gait assessments (e.g., normal, tandem, and dual-task gait) utilising force plates or inertial measurement units (IMU) were most consistent across time points extending to eight months. Outcome variables including step length, step time, and gait velocity were most reliable.

#### 2.3.3.5 Inter-rater reliability

Correlations for inter-rater reliability of non-instrumented assessments performed on healthy controls ranged from 0.20 to 0.99, with measurement property quality *adequate* for all studies. Static balance assessments included, BESS (n = 4) which ranged from 0.20-0.96 when using 3 assessors (197, 202, 203), and mBESS (n = 2), with reliability ranging from 0.80-0.83 using 2 and 3 assessors respectively (204, 205). Gait assessments included tandem gait (n = 1), TUG (n = 1), and walking on balance beam (n = 1). The TUG demonstrated best inter-rater reliability (ICC = 0.99) amongst two assessors. Other assessments consisted of a military-specific task; run-roll-aim (n = 1), with reliability ranging from 0.28-0.89 (206).

#### 2.3.3.6 Validity

The validity of 32 different assessments was reported across 35 studies; 17 studies assessed the validity of more than one test. Assessments were categorised into static balance (n = 21 studies, 13 different assessments), dynamic balance (n = 8 studies, 8 different assessments), gait (n =

13 studies, 8 different assessments), or other (n = 3 studies, 2 different assessments), and analysed either construct (n = 30) or known group validity (n = 8 studies). Studies were conducted on adults (n = 24) and minors (n = 11), with a total sample size of 1,417 concussed and 1616 control participants. A summary of study characteristics is presented in Table 2.2, 2.3, 2.4, and 2.5. Full table of study characteristics are presented in Supplementary Table 2.13 through to Table 2.20.

#### 2.3.3.7 Construct validity

##### 2.3.3.7.1 Static balance assessments

Outcome variables for non-instrumented static assessments included time or number of errors. Instrumented assessments reported, COM displacement, and COP displacement using force plates, IMUs, smartphones, or laboratory equipment. Time since concussion ranged from 24-hours to eight months with a tendency for *insufficient* sensitivity as time increased. Assessments included the BESS (n = 3), instrumented BESS (n = 2), balance accelerometry measure (BAM) (n = 1), mBESS (n = 7), instrumented mBESS (n = 4), SOT (n = 3), balance tracking system (n = 1), modified clinical test of sensory interaction in balance (MCTSIB) (n = 1), instrumented MCTSIB (n = 1), Phybrata system (n = 1), and virtual reality static balance (n = 1). On average, non-instrumented assessments, BESS and mBESS displayed *sufficient* sensitivity when conducted within 48-hours of sustaining a concussion (156, 207). However, sensitivity was *insufficient* when conducted beyond this period, and up to two months post-concussion (208). Instrumented BESS displayed *sufficient* sensitivity up to six months post-concussion (146). Virtual reality balance and Phybrata system displayed *sufficient* sensitivity at 10- and 30-days respectively and are a promising alternative to current assessments if equipment is made more readily available for clinicians (209, 210).

##### 2.3.3.7.2 Dynamic balance assessments

Outcome variables for non-instrumented assessments included time, heart rate, or number of errors. Instrumented assessments reported, COM displacement, COP displacement, or reach distance using force plates, IMUs or laboratory equipment. Time since concussion ranged from 24-hours to eight months with a tendency for *insufficient* sensitivity as time increased.

Assessments included: physical and neurological examination of subtle signs (PANESS) (n = 1), community balance and mobility scale (n = 2), Kasch pulse recovery test (KPR) (n = 1), instrumented Y balance test (YBT) (n = 1), battery assessments (n = 2), Computer-Assisted Rehabilitation ENvironment (CAREN) system (n = 1). The KPR test displayed *sufficient* sensitivity when conducted on adolescents (211). All assessments except for the battery assessments displayed *sufficient* sensitivity for adult populations. However, only the PANESS assessment reported time since concussion, with *sufficient* sensitivity up to 14-days post-concussion.

#### 2.3.3.7.3 Gait assessments

Outcome variables for non-instrumented gait assessments included time to complete or number of errors. Instrumented assessments provided more objective outcomes including COM displacement, step length, step time, cadence, anterior-posterior and medio-lateral accelerations, and gait velocity using pressure sensitive walkways, IMUs, smartphones, or other laboratory equipment. Time since concussion ranged from same day to three years with a tendency for *insufficient* sensitivity as time increased. Assessments included: functional gait assessment (n = 2), tandem gait (n = 5), complex tandem gait (n = 1), dual-task tandem gait (n = 3), dual-task gait (n = 1), instrumented gait (n = 3), instrumented dual-task gait (n = 3), and battery of gait assessments (n =1). In general, sensitivity remained *sufficient* up to two weeks for instrumented assessments, and seven days for non-instrumented assessments. Time to complete task was the primary outcome measure for non-instrumented assessments.

#### 2.3.3.7.4 Other assessments

Other assessments included a military-specific assessment, the Warrior Test of Tactile Agility (n = 1). This assessment was performed 2-years post-concussion and required participants to perform various motor tasks including forward/backward run, lateral shuffle, combat roll, and changes in position (e.g., lying to standing). The lowering and rolling movements within the assessment battery demonstrated *sufficient* AUC (0.83) (212).

Table 2.2 Overview of reliability, validity, and measurement error for static balance assessments for assessments used to monitor movement changes following a concussion.

| | Time taken | Equipment | Test procedure | Outcome variable | Reliability | Validity | Measurement error | Optimal conditions |
|---|---|---|---|---|---|---|---|---|
| Non-instrumented Static Assessments | | | | | | | | |
| *Balance Error Scoring System (BESS)* | | | | | | | | |
| (156, 198, 199, 202-204, 207, 208, 213, 214) | 2 trials, 4 conditions: 5 mins | Stopwatch<br>Foam pad | Double leg, single leg and tandem stance on firm surface and foam<br>Number of errors each 30-sec stance recorded | Number of errors | r = 0.41 to 0.94<br>Inter-tester: r = 0.20-0.96 | Sens = 0.23 to 0.60<br>Spec = 0.50 to 0.92<br>AUC = 0.51 to 0.63 | MDC test-retest = 7.3 errors<br>MDC inter-tester = 9.4 errors | 3 examiners, minimum of 2 trials on 4 conditions (excluding double leg stances) |
| *Modified Balance Error Scoring System (mBESS)* | | | | | | | | |
| (114, 151, 156, 198, 204, 205, 207, 208, 215-219) | ~2-5-minutes | Stopwatch | Double leg, single leg, tandem stance on firm surface only<br>Number of errors each 30-sec stance recorded | Number of errors | r = 0.02-0.88<br>Inter-tester: r = 0.80-0.83 | Sens = 0.05-0.71<br>Spec = 0.0.63-0.85<br>AUC = 0.54-0.72<br>Between group differences: p = <0.001-0.23 | - | Minimum of 2 trials<br>Familiarisation required for children |
| *Single Leg Stance (SLS)* | | | | | | | | |
| (220, 221) | ~2-5-minutes | Stopwatch | Errors recorded in 30-sec stance period | Number of errors | r = 0.85 | - | - | Most participants reached 30-sec cut off easily |
| *Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)* | | | | | | | | |
| (218) | ~5-10-minutes | NA | Double leg stance; stable surface eyes open, stable surface eyes closed, unstable surface eyes open, unstable surface eyes closed | Number of errors | - | Sens = 0.37<br>Spec = 0.88 | - | Poor sensitivity approximately 2 weeks post-concussion |
| *Paediatric Clinical Test of Sensory Interaction in Balance (PCTSIB)* | | | | | | | | |
| (218) | ~5-10-minutes | NA | Double leg stance; stable surface eyes open, stable surface eyes closed, unstable surface eyes open, unstable surface eyes closed | Number of errors | - | Between group differences: p = 0.46-0.49 | - | |

Table 2.2 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| Instrumented Static Assessments | | | | | | | | |
| *Instrumented Balance Error Scoring System* | | | | | | | | |
| (146, 197, 203, 208, 214) | ~5-minutes | Force plate or IMU | BESS performed on force plate (foam pad placed on plate) | Number of errors, COP or COM displacement | r = 0.88-0.89 | Sens = 0.38-0.0.94<br>Spec = 1<br>AUC = 0.70 | SEM = 0.00-0.45 | Perform in first month post-concussion to maximise sensitivity |
| *Instrumented Modified Balance Error Scoring System* | | | | | | | | |
| (145, 208, 219, 222, 223) | ~2-5-minutes | Force plate or IMU | MBESS on force plate or with IMU | Number of errors, COP or COM displacement | Force plate: r = 0.69-0.77<br>Smartphone: r = 0.66-0.75 | Sens = 0.42-0.92<br>Spec = 0.76 - 1.00<br>AUC = 0.74-0.91 | - | |
| *Instrumented Single Leg Stance (Instrumented SLS)* | | | | | | | | |
| (224) | ~2-5-minutes | Force plate | Measure of COP on force plate<br>Eyes open and eyes closed conditions | Number of errors, COP displacement | r = 0.58-0.88 | | SEM = 0.41-2.97 | Eyes closed condition displayed better reliability |
| *Balance Accelerometry Measure (BAM)* | | | | | | | | |
| (190, 225, 226) | ~5-10-minutes | IMU, smartglasses, or smartphone | 6 stances: firm surface - double leg eyes open, double leg eyes closed, tandem eyes open, tandem eyes closed<br>Foam surface – double leg eyes open, double leg eyes closed<br>Number of errors each 45-sec recorded | Number of errors, COM displacement | r = 0.28-0.86 | AUC <2 weeks = 0.74-0.77<br>AUC >2 weeks = 0.67-0.68 | - | Perform within 2 weeks post-concussion |
| *Balance Tracking System* | | | | | | | | |
| (196, 227) | ~2-minutes | Force plate | 4 x 20-sec trials: 1 familiarisation and 3 recorded. Participant standing on force plate with eyes closed and hands on hips. COP recorded | COP displacement | - | Sens = 0.64 | - | Baseline scores better for comparison |

Table 2.2 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| *SWAY Balance* | | | | | | | | |
| (228) | ~2-5-minutes | Smartphone | 5 x 10-sec stances: bipedal, tandem (R), tandem (L), single leg (R), single leg (L) on firm surface with eyes closed. SWAY scores calculated | COM displacement | r = 0.47-0.86 | - | SEM = 5.77-7.56 | 3 sessions<br>Mobile held on sternum |
| *Sensory Organisation Test (SOT)* | | | | | | | | |
| (229-232) | ~10-20-minutes | SOT laboratory equipment | 6, 20s conditions: stance eyes open, eyes closed, eyes open; visual interference, eyes open; surface interference, eyes closed; surface interference, eyes open; visual and surface interference. | COP displacement | r = 0.3-0.72 | Sens = 0.13-0.73<br>Spec = 0.85-0.95<br>AUC = 0.79 | SEM = 4.92-5.00 | |
| *Physiological vibration acceleration (Phybrata) System* | | | | | | | | |
| (209) | ~2-minutes | IMU | 20-second stance: eyes open and eyes closed | COM displacement | - | Sens = 0.92<br>Spec = 0.94<br>AUC = 0.98 | - | |
| *Instrumented Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)* | | | | | | | | |
| (233) | ~5-10-minutes | Force plate | Double leg stance; stable surface eyes open and eyes closed, unstable surface eyes open and eyes closed | Number of errors, COP displacement | - | Sens = 0.37<br>Spec = 0.88 | - | |
| *Virtual Reality Balance* | | | | | | | | |
| (210, 234) | ~10-minutes | Force plate<br>Virtual reality system | Tandem stance, hands on hips. Participant placed in harness in 3D virtual reality room. 10 trials performed, 1 with the room still, 9 with it moving. Each trial lasted 30-seconds | COP displacement | - | Sens = 0.86<br>Spec = 0.88<br>AUC = 86<br>Between group differences: p = 0.006-<0.001 | - | 10 trials – 1 trial with still room, 9 trials with rotating room |

Table 2.3 Overview of reliability, validity and measurement error for dynamic motor assessments used to monitor movement changes following a concussion.

| | Time taken | Equipment | Test procedure | Outcome variable | Reliability | Validity | Measurement error | Optimal conditions |
|---|---|---|---|---|---|---|---|---|
| Non-instrumented Dynamic Assessments | | | | | | | | |
| *Clinical Reaction Time* | | | | | | | | |
| (213) | ~3-minutes | Pole | Stick drop test: converted to speed | Distance on stick, speed | r = 0.32 | - | - | |
| *Kasch Pulse Recovery (KPR)* | | | | | | | | |
| (211) | ~5-minutes | NA | 3-minute stair climb at rate of 24-steps.min Exercise HR and HR recovery (1-minute seated) recorded | HR pre, during, and post | - | Sens = 1.00<br>Spec = 0.96<br>AUC = 0.98 | - | Performed on children |
| *Physical and Neurological Examination of Subtle Signs (PANESS)* | | | | | | | | |
| (235) | ~10-15-minutes | Stopwatch | Gait: heel walk, toe walk, walk on side of feet, tandem gait Stations: 20-sec stance (a) feet together, eyes closed, arms out, (b) eyes closed, tongue protruding, (c) tandem stance, hop each foot (50 hops) | Number of errors | - | Sens = 0.76<br>Spec = 0.90 | - | |
| Instrumented Dynamic Assessment | | | | | | | | |
| *Instrumented Y Balance Test (Instrumented YBT)* | | | | | | | | |
| (200, 236) | ~5-minutes | IMU | Standing on 1 leg, tap toe of non-standing leg in anterior, posterior-medial, and posterior lateral directions | Reach distance, COM displacement | r = 0.76-0.99 | Sens = 0.76<br>Spec = 0.53 | - | Improved validity with use of lumbar IMU |

Table 2.3 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| *Instrumented Limits of Stability Test (Instrumented LOS)* | | | | | | | | |
| (201, 237) | ~5-10-minutes | Force plate | 8 trials shifting centre of gravity in directions: forward, right/forward, right, right/back, back, left/back, left, left/forward | COP displacement | r = 0.973-0.96 | - | SEM = 0.35-1.17 | |
| *Dynamic Postural Stability Index (DPSI) and DPSI with Dual-Task* | | | | | | | | |
| (224) | ~2-5-minutes | Force plate | 3 trials of jumps: forward, lateral, and 90º rotation | COP during landing | DPSI: r = 0.58-0.93<br>DPSI DT: r = 0.32-0.80 | - | DPSI SEM = 0.0047-0.023<br>DPSI+DT SEM = 0.004-0.019 | Improved reliability with standard DPSI |
| *Postural Stress Test (PST)* | | | | | | | | |
| (238) | ~5-10-minutes | Pulley system | Pulley system providing external perturbations. Participant required to maintain upright position. Amount of weight required to cause stepping postural adjustment is recorded | Amount of weight required to maintain balance | - | Between group differences: p = 0.027-0.024 | - | |
| *Computer-Assisted Rehabilitation ENvironment (CAREN) System* | | | | | | | | |
| (239) | | Accelerometry, motion capture | Virtual reality standing and walking with external perturbations | COM displacement | - | Sens = 0.65-0.90 | - | Walking with perturbations more sensitive<br>10 trials performed |
| *Battery combined: Grip strength, single leg jump, MCTSIB* | | | | | | | | |
| (233) | ~10-15-minutes | Force plate, tape measure, handheld dynamometer | Grip strength using handheld dynamometer. Single leg standing long jump. MCTSIB: Double leg stance; stable surface eyes open, stable surface eyes closed, unstable surface eyes open, unstable surface eyes closed | Number of errors, COP displacement, grip strength | - | Sens = 0.41<br>Spec = 0.77 | - | Individual measures of MCTSIB, grip strength, and standing long jump displayed poor sensitivity. Higher sensitivity with combined score |

Table 2.3 continued

| | | | | | | | |
|---|---|---|---|---|---|---|---|
| *Bruininks-Oseretsky Test of Motor Proficiency* | | | | | | | |
| (238) | 15-60-minutes total<br>Balance component includes 8 items ~15-minutes | Requires specialised equipment | Gross motor: running speed and agility, balance, bilateral coordination, strength<br>Fine motor: response speed, visual motor control, upper-limb dexterity | Number of errors, time to complete trials | - | Between group differences: p = 0.024-0.001 | - |

Table 2.4 Overview of reliability, validity and measurement error for gait assessments used to monitor movement changes following a concussion.

| | Time taken | Equipment | Test procedure | Outcome variable | Reliability | Validity | Measurement error | Optimal Conditions |
|---|---|---|---|---|---|---|---|---|
| Non-instrumented Gait Assessments | | | | | | | | |
| *Community Balance and Mobility Scale* | | | | | | | | |
| (240, 241) | ~10-15-minutes | Stopwatch, rigid box, 1-3-kg weight | Tasks: unilateral stance, tandem walk, 180° tandem pivot, lateral foot scooting, hopping forward, crouch and walk, lateral dodging, walk and look, run with stop, forward/backward walk, walk, look and carry, descend stairs, step-ups<br>Scoring on scale 0-5 | Number of errors, time to complete | - | Sens = 0.78-1.00<br>Spec = 0.88-0.91<br>AUC = 0.92-0.98 | - | Performed on adult population<br>Perform within 2 weeks post-concussion |
| *Timed Up and Go (TUG)* | | | | | | | | |
| (242) | 1-2-minutes | Chair<br>Tape measure | Seated in chair, participant stands and walks 3m, turns 180° and sits back down | Time to complete, errors | r = 0.85<br>Inter-tester: r = 0.99 | - | - | |
| *Walking on Balance Beam* | | | | | | | | |
| (242) | 1-2-minutes | Balance beam, stopwatch | Participants walk on balance beam | Number of errors | r = 0.87<br>Inter-tester: r = 0.35 | - | - | |

Table 2.4 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| *Tandem Gait* | | | | | | | | |
| (114, 151, 156, 215, 217, 220, 221, 242-244) | 2-5-minutes | Stopwatch, measuring tape | Heel to toe walking | Time to complete, number of errors | r = 0.46-0.98<br>Inter-tester: r = 0.70 | Sens = 0.63-0.88<br>Spec = 0.61-0.72<br>AUC = 0.55-0.86<br>Between group differences: p = <0.001 | SEM = 0.07 | |
| *Dual-Task Tandem Gait* | | | | | | | | |
| (114, 151, 243, 244) | 2-5-minutes | Stopwatch, measuring tape | Heel and toe walk with addition of cognitive task (count back in 7's, recite alphabet backwards etc.) | Time to complete, number of motor and cognitive errors | r = 0.84-0.92 | Sens = 0.85<br>Spec = 0.72<br>AUC = 0.80-0.87<br>Between group differences: p = 0.002-<0.001 | SEM = 0.06 | Cognitive tasks need to be age-specific |
| *Complex Tandem Gait* | | | | | | | | |
| (218) | 2-5-minutes | Stopwatch, measuring tape | Normal gait with addition of cognitive task (count back in 7's, recite alphabet backwards etc.) | Time to complete, number of errors | - | Sens = 0.41<br>Spec = 0.90 | - | Cognitive tasks need to be age-specific |
| *Gait* | | | | | | | | |
| (151) | 2-5-minutes | Stopwatch, measuring tape | Normal gait | Time to complete | - | Between group differences: p = 0.006 | - | |
| *Dual-Task Gait* | | | | | | | | |
| (151, 245) | 2-5-minutes | Stopwatch, measuring tape | Normal gait with addition of cognitive task (count back in 7's, recite alphabet backwards etc.) | Time to complete, number of motor and cognitive errors | - | Sens = 0.77<br>Spec = 1.00<br>Between group differences: p = 0.29 | - | Cognitive tasks need to be age-specific |

Table 2.4 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| Instrumented Gait Assessments | | | | | | | | |
| *Battery combined: Berg balance, Balance Evaluation Systems Test (BESTest), Dynamic gait index, High-Level Mobility Assessment Tool (HiMat)* | | | | | | | | |
| (239) | ~60-minutes | HiMat, tape measure, step, chair, ramp, obstacle | Berg balance: sit to stand, stand, sit, stand to sit, transfer, stand with eyes closed, stand with feet together, reach forward, reach to floor, turn to look behind, turn 360º, step, tandem stance, single leg stance<br>BESTest: 36 items assessing bio-mechanical constraints, stability limits, anticipatory response, postural response, sensory orientation, stability in gait<br>DGI: 8 items – walking on level surface, changing speed, head turn horizontal, head turn vertical, 180º turn, step over, step around, stair ascent, descent.<br>HiMat: 13 items – walk, walk backward, walk on toes, over obstacle, run, skip, hop, bound, stair ascent, descent | Number of errors, time to complete | - | Score ‘no impairment’ = 89% concussed | - | 10 trials |

Table 2.4 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| *Battery Combined: gait, tandem gait, dual-task tandem gait* | | | | | | | | |
| (151) | ~10-20-minutes | Smartphone | Gait: assessing cadence and velocity (instrumented). Tandem gait: time to complete (non-instrumented) Dual-task tandem gait: time to complete (non-instrumented) | Time to complete, COM displacement | - | AUC = 0.91 | SEM = 0.05 | Combined score displayed highest AUC |
| *Instrumented Dual-Task Tandem Gait* | | | | | | | | |
| (246, 247) | 2-5-minutes | IMU | Heel to toe walk with addition of cognitive task (count back in 7's, recite alphabet backwards etc.) | COM displacement, accelerometer metrics, number of cognitive errors | r = 0.73-0.94 | - | SE = -0.080-1.355 | Cognitive tasks need to be age-specific |
| *Instrumented Gait* | | | | | | | | |
| (113, 115, 151, 246-252) | 2-5-minutes | IMU, 3D motion capture, GAITRite, smartphone, virtual reality | Normal gait | Time to complete, COM or COP displacement | r = 0.10-0.99 Internal consistency: r = 0.76-0.97 | Sens = 0.26-0.63 Spec = 0.62-0.85 AUC = 0.76-0.79 Between group differences: p = 0.99-0.002 | MDC = 0.875-7.42 SEM = -1.32-0.08 | |
| *Instrumented Dual-Task Gait* | | | | | | | | |
| (113, 115, 151, 251-253) | 2-5-minutes | IMU or 2D motion analysis | Normal gait with addition of cognitive task (count back in 7's, recite alphabet backwards etc.) | COM or COP displacement, time to complete, number of motor and cognitive errors | Internal consistency: r = 0.61-0.97 | Sens = 0.26-0.70 Between group differences: p = 0.93-0.07 | MDC = 0.839-6.76 | Cognitive tasks need to be age-specific |

Table 2.4 continued

| | | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| *Functional Gait Assessment* | | | | | | | | |
| (240, 241) | 5-20-minutes | Stopwatch, measuring tape, 22cm high obstacle, stairs | 10 items; 6-m gait, 6-m gait alternating speed, gait with horizontal head turns, gait with vertical head turns, gait and pivot turn, step over obstacle, gait with narrow base of support, gait with eyes closed, backwards gait, ascend/descend stairs | Time to complete, number of errors | - | Sens = 0.05-0.88<br>Spec = 0.75-1.00 | - | Performed on adult population |

Table 2.5 Overview of reliability, validity and measurement error for other motor assessments used to monitor movement changes following a concussion.

| | Time taken | Equipment | Test procedure | Outcome variable | Reliability | Validity | Measurement error | Optimal conditions |
|---|---|---|---|---|---|---|---|---|
| Non-instrumented Task Specific Assessment | | | | | | | | |
| *Run-Roll-Aim* | | | | | | | | |
| (206) | 10-20-minutes | NA | 4 trials, military course with combat specific tasks including; roll and aim | Time to complete, number of errors | Inter-tester: r = 0.28-0.89 | Between group differences: p = <0.01 | - | Military specific |
| *Portable Warrior Test of Tactile Agility* | | | | | | | | |
| (254) | 10-20-minutes | Stopwatch | 5 trials, military course with combat specific tasks including; run, roll, and backwards run. Performed under single-task and dual-task conditions | Time to complete, number of errors | - | Between group differences single-task: p = <0.001<br>Between group differences dual-task: p = 0.004 | - | Military specific |
| Instrumented Task Specific Assessment | | | | | | | | |
| *Instrumented Portable Warrior Test of Tactile Agility* | | | | | | | | |
| (212) | 10-20-minutes | IMU | 5 trials, military course with combat specific tasks including; run, roll, and backwards run. Performed under single-task and dual-task conditions | Time to complete, number of errors, COM or COP displacement | - | AUC of 'lowering and rolling' movements: AUC = 0.83<br>Between group differences: 0.08 - <0.0001 | - | Military specific |

Table 2.6 Test-retest of static balance assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Balance Error Scoring System* | | | | | | |
| Broglio et al., 2009 | n = 48 | ~20 years | ~50 days | Adequate | 2 trials: G = 0.82-0.85 | + |
| Hunt et al., 2009 | n = 78 | ~16 years | NA | Adequate | 1-7 trials: r = 0.60-0.94 | -/+ |
| Chang et al., 2014 | n = 30 | ~24 years | 7 days | Adequate | r = 0.61-0.78 | + |
| Broglio et al., 2018 | n = 4874 | ~19 years | ~11 months | Adequate | r = 0.41 | - |
| | | | ~ 20 months | | r = 0.42 | - |
| Finnoff et al., 2009 | n = 30 | | 2 days | Adequate | r = 0.57 | - |
| Alsalaheen et al., 2015 | n = 36 | ~15 years | 7 days | Adequate | r = 0.13-0.74 | -/+ |
| *Instrumented Balance Error Scoring System* | | | | | | |
| Chang et al., 2014 | n = 30 | ~24 years | 7 days | Adequate | Force plate: r = 0.89 | + |
| | | | | | WBB: r = 0.88 | + |
| Alsalaheen et al., 2015 | n = 36 | ~15 years | 7 days | Adequate | r = 0.19-0.74 | -/+ |
| *Modified Balance Error Scoring System* | | | | | | |
| Hunt et al., 2009 | n = 144 | ~16 years | NA | Adequate | 3 trials: r = 0.73-0.88 | + |
| Nelson et al., 2017 | n = 57 | 9-13 years | ~2 months | Adequate | 2 trials: r = 0.02 | - |
| Kontos et al., 2021 | n = 108 | 18-40 years | ~ 6 months | Adequate | r = 0.59 | - |
| *Instrumented Modified Balance Error Scoring System* | | | | | | |
| German et al., 2022 | n = 28 | ~28 years | Same day | Adequate | Force plate: r = 0.69-0.77 | -/+ |
| | | | | | Phone: r = 0.66-0.75 | -/+ |
| *Single Leg Stance* | | | | | | |
| Schneiders et al., 2008 | n = 30 | ~21 years | 15 mins | Doubtful | r = 0.85 | + |
| Schneiders et al., 2010 | n = 172 | ~22 years | ~1 week | Adequate | r = 0.85 | + |
| *Instrumented single leg stance* | | | | | | |
| Westwood et al., 2020 | n = 12 | ~22 years | 10 days | Adequate | Eyes Open | |
| | | | | | r = 0.58-0.84 | - |
| | | | | | Eyes Closed | + |
| | | | | | r = 0.77-0.88 | + |
| | | | | | ML r = 0.77 | |
| | | | | | AP r = 0.87 | |
| | | | | | Vertical r = 0.88 | |
| *Balance Accelerometry Measure* | | | | | | |
| Marchetti et al., 2013 | n = 84 | ~47 years | 15 mins | Adequate | r = 0.74-0.86 | + |
| Salisbury et al., 2018 | n = 42 | ~24 years | Same day | Doubtful | r = 0.28-0.82 | -/+ |

Table 2.6 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Balance tracking system* | | | | | | |
| Hearn et al., 2018 | n = 20 | ~23 years | 1, 3, 8, 15 days | Adequate | r = 0.92 | + |
| *SWAY Balance mobile application* | | | | | | |
| Amick et al., 2015 | n = 24 | ~26 years | 1 week | Adequate | r = 0.47 | - |
| | | | 2 weeks | | r = 0.78 | + |
| | | | 3 weeks | | r = 0.75 | + |
| *Sensory Organisation Test* | | | | | | |
| Register et al., 2013 | n = 38 | ~21 years | ~1 month | Adequate | r = 0.3 | - |
| Christy et al., 2019 | n = 87 | ~21 years | 7-12 months | Adequate | r = 0.72 | + |

Table 2.7 Inter-tester reliability of static balance assessments

| | Participants | Age | No of Assessors | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Balance Error Scoring System* | | | | | | |
| Finnoff et al., 2009 | n = 30 | >18 years | n = 3 | Adequate | r = 0.57 | - |
| Chang et al., 2014 | n = 30 | ~24 years | n = 3 | Adequate | Total r = 0.20-0.35 | - |
| | | | | | Individual r = 0.00-0.69 | - |
| Riemann et al., 1999 | n = 111 | ~20 years | n = 18 | Adequate | Stance variations: r = 0.65-0.96 | -/+ |
| Kleffelgaard et al., 2018 | n = 42 | ~23 years | n = 2 | Adequate | r = 0.66 | - |
| *Modified Balance Error Scoring System* | | | | | | |
| Kleffelgaard et al., 2018 | n = 42 | ~23 years | n = 2 | Adequate | r = 0.80 | + |
| Glass et al., 2019 | n = 25 | ~25 years | n = 3 | Adequate | r = 0.83 | + |

Table 2.8 Test-retest reliability of dynamic balance assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Instrumented Y Balance Test* | | | | | | |
| Johnston et al., 2017 | n = 15 | ~23 years | Same day | Adequate | r = 0.76-0.99 | + |
| *Clinical Reaction Time* | | | | | | |
| Broglio et al., 2018 | n = 4874 | ~20 years | ~11 months | Adequate | r = 0.32 | - |

Table 2.8 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Instrumented Limits of Stability Test* | | | | | | |
| Lininger et al., 2018 | n = 27 | ~24 years | ~7 days | Adequate | r = 0.95-0.96 | + |
| Alsalaheen et al., 2015 | n = 15 | ~16 years | 7 days | Adequate | r = 0.73-0.96 | + |
| *Dynamic Postural Stability Index (DPSI) and DPSI with Dual-Task* | | | | | | |
| Westwood et al., 2020 | n = 12 | ~22 years | 10 days | Adequate | DPSI r = 0.58-0.93 | -/+ |
| | | | | | DPSI DT r = 0.32-0.80 | -/+ |

Table 2.9 Test-retest reliability of gait-based motor assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Tandem Gait* | | | | | | |
| Schneiders et al., 2008 | n = 30 | ~20 years | Same day | Doubtful | r = 0.98 | + |
| Schneiders et al., 2010 | n = 172 | ~22 years | ~7 days | Adequate | r = 0.97 | + |
| Eemanipure et al., 2012 | n = 133 | ~25 years | 7 days | Adequate | r = 0.95 | + |
| Nelson et al., 2017 | n = 155 | 5-13 years | ~2-4 months | Adequate | r = 0.46 | - |
| Howell et al., 2019 | n = 32 | ~14 years | 2, 4 weeks | Adequate | r = 0.86 | + |
| Wingerson et al., 2020 | n = 44 | ~15 years | ~7 days | Doubtful | r = 0.93 | + |
| *Instrumented Gait* | | | | | | |
| Howell et al., 2020 | n = 20 | ~22 years | ~2 weeks | Adequate | r = 0.86-0.95 | + |
| Howell et al., 2017 | n = 265 | ~20 years | ~8 months | Adequate | r = 0.68-0.80 | -/+ |
| Nishinguchi et al., 2012 | n = 30 | ~20 years | Same day | Adequate | r = 0.75-0.91 | + |
| Kuznetsov et al., 2018 | n = 32 | ~25 years | ~7 days | Adequate | r = 0.59-0.96 | -/+ |
| Silsupadol et al., 2017 | n = 12 | ~22 years | Same day | Doubtful | r = 0.70-0.99 | + |
| | n = 22 | ~74 years | | | | |
| Howell et al., 2021 | n = 17 | ~17 years | ~1 month | Adequate | r = 0.10-0.94 | -/+ |
| Wilkerson et al., 2021 | n = 9 | ~25 years | ~3 months | Adequate | r = 0.83-0.97 | + |
| *Dual-Task Tandem Gait* | | | | | | |
| Howell et al., 2019 | n = 32 | ~15 years | 2, 4 weeks | Adequate | r = 0.84 | + |
| Wingerson et al., 2020 | n = 44 | ~15 years | ~7 days | Doubtful | r = 0.92 | + |
| *Instrumented Dual-Task Tandem Gait* | | | | | | |
| Howell et al., 2020 | n = 20 | ~22 years | ~2 weeks | Adequate | r = 0.80-0.94 | + |
| Howell et al., 2017 | n = 265 | ~20 years | ~8 months | Adequate | r = 0.73-0.85 | + |

Table 2.9 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Instrumented Dual-Task Gait* | | | | | | |
| Howell et al., 2021 | n = 17 | ~17 years | ~1 month | Adequate | r = 0.61-0.94 | -/+ |
| Wilkerson et al., 2021 | n = 9 | ~25 years | ~3 months | Adequate | r = 0.81-0.97 | + |
| *Timed Up and Go* | | | | | | |
| Eemanipure et al., 2012 | n = 133 | ~25 years | 7 days | Adequate | r = 0.85 | + |
| *Walking on Balance Beam* | | | | | | |
| Eemanipure et al., 2012 | n = 133 | ~25 years | 7 days | Adequate | r = 0.87 | + |

Table 2.10 Inter-tester reliability of gait-based motor assessments

| | Participants | Age | No of Assessors | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Tandem Gait* | | | | | | |
| Eemanipure et al., 2012 | n = 133 | ~25 years | n = 2 | Adequate | r = 0.70 | + |
| *Timed Up and Go* | | | | | | |
| Eemanipure et al., 2012 | n = 133 | ~25 years | n = 2 | Adequate | r = 0.99 | + |
| *Walking on Balance Beam* | | | | | | |
| Eemanipure et al., 2012 | n = 133 | ~25 years | n = 2 | Adequate | r = 0.35 | - |

Table 2.11 Internal consistency of gait-based motor assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Instrumented Gait* | | | | | | |
| Howell et al., 2016 | n = 24<br>n = 21 | ~15 years<br>~22 years | 1 week<br>2 weeks<br>1 month<br>2 months | Inadequate | r = 0.76-0.97 | + |
| *Instrumented Dual-Task Gait* | | | | | | |
| Howell et al., 2016 | n = 24<br>n = 21 | ~15 years<br>~22 years | 1 week<br>2 weeks<br>1 month<br>2 months | Inadequate | r = 0.79-0.97 | + |

Table 2.12 Inter-tester reliability of task-specific assessments

| | Participants | Age | No of Assessors | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Run-Roll-Aim* | | | | | | |
| Prim et al., 2019 | n = 50 | ~33 years | n = 2 | Adequate | r = 0.28-0.89 | -/+ |

Table 2.13 Discriminant validity of static balance assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Balance Error Scoring System* | | | | | | |
| Buckley et al., 2018 | n = 35 | ~18 years | <24 hours | Adequate | Sensitivity = 0.60 | - |
| Oldham et al., 2018 | n = 38 | ~20 years | <48 hours | Very Good | Sensitivity = 0.45 | - |
| | n = 38 | | | | Specificity = 0.50 | - |
| | control | | | | AUC = 0.51 | - |
| King et al., 2014 | n = 13 | ~16 years | 2-5 months | Very Good | Sensitivity = 0.23 | - |
| | n = 13 | | | | Specificity = 0.92 | + |
| | control | | | | AUC = 0.63 | - |
| *Instrumented Balance Error Scoring System* | | | | | | |
| King et al., 2014 | n = 13 | ~16 years | | Very Good | Sensitivity = 0.38 | - |
| | n = 13 | | | | Specificity = 1 | + |
| | control | | | | AUC = 0.70 | + |
| Pryhoda et al., 2020 | n = 25 | >18 years | 3 days | Adequate | r = 0.69-0.94 | -/+ |
| | n = 92 | | 1 week | | r = 0.53-0.84 | -/+ |
| | control | | 1 month | | r = 0.41-0.92 | -/+ |
| | | | 6 months | | r = 0.60-0.80 | -/+ |
| *Balance Accelerometry Measure (BAM)* | | | | | | |
| Furman et al., 2013 | n = 43 | ~16 years | <2 weeks | Very Good | AUC = 0.74-0.77 | + |
| | n = 27 | | >2weeks | | AUC = 0.67-0.68 | - |
| | control | | | | | |

Table 2.13 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Modified Balance Error Scoring System* | | | | | | |
| VanDeventer et al., 2021 | n = 81<br>n = 58<br>control | ~15 years | | Very Good | AUC = 0.71 | + |
| Hanninen et al., 2018 | n = 27<br>n = 179<br>control | ~28 years | Day of injury | Adequate | AUC = 0.69-0.72 | -/+ |
| Corwin et al., 2020 | n = 78<br>n = 88<br>control | ~16 years | ~2 weeks | Very Good | Sensitivity = 0.05-0.60 | - |
| Buckley et al., 2018 | n = 35 | ~18 years | <24 hours<br>RTP | Adequate | Sensitivity = 0.71<br>Sensitivity = 0.65 | +<br>- |
| Oldham et al., 2018 | n = 38<br>n = 38<br>control | ~20 years | <48 hours | Very Good | Sensitivity = 0.45<br>Specificity = 0.63<br>AUC = 0.54 | -<br>-<br>- |
| *Modified Balance Error Scoring System cont.* | | | | | | |
| King et al., 2017 | n = 52<br>n = 76<br>control | ~20 years | ~3 days | Very Good | Sensitivity = 0.35<br>AUC = 0.61 | -<br>- |
| King et al., 2014 | n = 13<br>n = 13<br>control | ~16 years | ~ | Very Good | Sensitivity = 0.31<br>Specificity = 0.85<br>AUC = 0.64 | -<br>+<br>- |
| *Instrumented Modified Balance Error Scoring System* | | | | | | |
| King et al., 2017 | n = 52<br>n = 76 | ~20 years | ~3 days | Very Good | Sensitivity = 0.59-0.63<br>AUC = 0.74-0.75 | -<br>+ |
| Doherty et al., 2017 | n = 15<br>n = 15<br>control | ~22 years | | Very Good | Sensitivity = 0.72-0.92<br>Specificity = 0.79-0.96<br>AUC = 0.91 | +<br>+<br>+ |
| King et al., 2014 | n = 13<br>n = 13<br>control | ~16 years | ~2-5 months | Very Good | Sensitivity = 0.54<br>Specificity = 1<br>AUC = 0.81 | -<br>+<br>+ |
| Baracks et al., 2018 | n = 48<br>n = 45<br>control | ~20 years | ~ 3 days | Very Good | Sensitivity = 0.42-0.52<br>Specificity = 0.76-0.80 | -<br>+ |

Table 2.13 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Sensory Organisation Test* | | | | | | |
| Register et al., 2013 | n = 132<br>n = 38<br>control | ~18 years | ~2 days | Very Good | Sensitivity = 0.13<br>Specificity = 0.95 | -<br>+ |
| Resch et al., 2016 | n = 40<br>n = 40<br>control | ~19 years | 24 hours | Adequate | Sensitivity = 0.73<br>Specificity = 0.85<br>AUC = 0.79 | +<br>+<br>+ |
| Broglio et al., 2008 | n = 63<br>n = 66<br>control | ~20 years | 24 hours<br>8 months | Very Good | Sensitivity = 0.29 | - |
| *Balance Tracking System* | | | | | | |
| Goble et al., 2016 | n = 25 | ~20 years | 48 hours | Adequate | Sensitivity = 0.64 | - |
| *Instrumented Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)* | | | | | | |
| Toong et al., 2021 | n = 32<br>n = 32<br>control | ~14 years | ~3 months | Very Good | Sensitivity = 0.03<br>Specificity = 0.93 | -<br>+ |
| *Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)* | | | | | | |
| Corwin et al., 2020 | n = 78<br>n = 88<br>control | ~16 years | ~2 weeks | Very Good | Sensitivity = 0.37<br>Specificity = 0.88 | -<br>+ |
| *Phybrata System (Instrumented)* | | | | | | |
| Ralston et al., 2020 | n = 92<br>n = 83<br>control | ~24 years | ~30 days | Very Good | Sensitivity = 0.92<br>Specificity = 0.94<br>AUC = 0.98 | +<br>+<br>+ |
| *Virtual Reality Balance* | | | | | | |
| Teel et al., 2016 | n = 27<br>n = 94<br>control | ~20 years | ~7-10 days | Doubtful | Sensitivity = 0.86<br>Specificity = 0.88<br>AUC = 0.86 | +<br>+<br>+ |

Table 2.14 Discriminant validity of dynamic balance assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Physical and Neurological Examination of Subtle Signs (PANESS)* | | | | | | |
| Stephens et al., 2020 | n = 20<br>n = 17 control | ~15 years | ~4-14 days | Adequate | Sensitivity = 0.76<br>Specificity = 0.90 | +<br>+ |
| *Community Balance and Mobility Scale* | | | | | | |
| Pape et al., 2016 | n = 8<br>n = 8 control | ~31 years | ~7 days | Adequate | Sensitivity = 1.00<br>Specificity = 0.88<br>AUC = 0.98 | +<br>+<br>+ |
| Pape et al., 2020 | n = 45<br>n = 45 control | ~33 years | | Adequate | Sensitivity = 0.78<br>Specificity = 0.91<br>AUC = 0.92 | +<br>+<br>+ |
| *Kasch Pulse Recovery (KPR)* | | | | | | |
| Fyffe et al., 2020 | n = 45 | ~13 years | | Very Good | Sensitivity = 1.00<br>Specificity = 0.96<br>AUC = 0.98 | +<br>+<br>+ |
| *Instrumented Y Balance Test* | | | | | | |
| Johnston et al., 2019 | n = 21<br>n = 88 control | ~22 years | | Adequate | Sensitivity = 0.76<br>Specificity = 0.53 | +<br>- |
| *Battery Assessments* | | | | | | |
| Toong et al., 2021 | n = 32<br>n = 32 control | ~13 years | ~1 month | Very Good | Sensitivity = 0.41<br>Specificity = 0.77 | -<br>+ |
| Rao et al., 2020 | n = 11<br>n = 10 control | ~39 years | | Adequate | Score of 'no impairment' = 89% MTBI | - |
| *CAREN System* | | | | | | |
| Rao et al., 2020 | Rao et al., 2020 | Rao et al., 2020 | Rao et al., 2020 | Rao et al., 2020 | Rao et al., 2020 | Rao et al., 2020 |

Table 2.15 Discriminant validity of gait-based motor assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Functional Gait Assessment and Gait Speed* | | | | | | |
| Pape et al., 2020 | n = 45<br>n = 45 control | ~33 years | | Adequate | Sensitivity = 0.05-0.69<br>Specificity = 0.76-1.00 | -/+<br>+ |
| Pape et al., 2016 | n = 8<br>n = 8 control | ~31 years | ~7 days | Adequate | Sensitivity = 0.75-0.88<br>Specificity = 0.75 | +<br>+ |
| *Tandem Gait* | | | | | | |
| Hanninen et al., 2018 | n = 27<br>n = 127 control | ~27 years | Day of injury | Adequate | AUC = 0.55-0.56 | - |
| Oldham et al., 2018 | n = 38<br>n = 38 control | ~20 years | <48 hours | Very Good | Sensitivity = 0.63<br>Specificity = 0.61<br>AUC = 0.70 | -<br>-<br>+ |
| VanDeventer et al., 2021 | n = 81<br>n = 58 control | ~14 years | | Very Good | Sensitivity = 0.88<br>Specificity = 0.72<br>AUC = 0.86 | +<br>+<br>+ |
| Wingerson et al., 2020 | n = 73<br>n = 44 control | ~15 years | ~7 days | Very Good | AUC = 0.85 | + |
| Howell et al., 2019 | n = 15<br>n = 31 control | ~16 years | ~7 days | Very Good | AUC = 0.86 | + |
| *Complex Tandem Gait* | | | | | | |
| Corwin et al., 2020 | n = 78<br>n = 88 control | ~16 years | ~2 weeks | Very Good | Sensitivity = 0.41<br>Specificity = 0.90 | -<br>+ |

Table 2.15 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Dual-Task Tandem Gait* | | | | | | |
| VanDeventer et al., 2021 | n = 81<br>n = 58<br>control | ~14 years | | Very Good | Sensitivity = 0.85<br>Specificity = 0.72<br>AUC = 0.84 | +<br>+<br>+ |
| Wingerson et al., 2020 | n = 73<br>n = 44<br>control | ~15 years | ~7 days | Very Good | AUC = 0.80 | + |
| Howell et al., 2019 | n = 15<br>n = 31<br>control | ~16 years | ~7 days | Very Good | AUC = 0.87 | + |
| *Instrumented Dual-Task Gait* | | | | | | |
| Howell et al., 2015 | n = 10<br>n = 7<br>control | ~19 years | 72 hours<br>1 week<br>2 weeks<br>1 month<br>2 months | Very Good | Sensitivity = 0.70<br>Sensitivity = 0.70<br>Sensitivity = 0.60<br>Sensitivity = 0.30<br>Sensitivity = 0.40 | +<br>+<br>-<br>-<br>- |
| Howell et al., 2019 | n = 54<br>n = 60<br>control | ~20 years | ~5 days | Adequate | Sensitivity =0.26-0.56 | - |
| Wilkerson et al., 2021 | n = 9<br>n = 7<br>control | ~25 years<br>~20 years | ~3 ± 2.2 years | Very Good | Reaction time dispersion (ST and DT combined)<br>Sensitivity = 0.89<br>Specificity = 0.86 | + |
| *Battery Gait Assessment* | | | | | | |
| Howell et al., 2019 | n = 15<br>n = 31<br>control | ~16 years | ~7 days | Very Good | AUC = 0.91 | + |

Table 2.16 Discriminant validity of task-specific motor assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Instrumented Portable Warrior Test of Tactile Agility* | | | | | | |
| Favorov er al., 2021 | n = 42<br>n = 57 control | ~29 years<br>~29 years | ≤ 2 years | Very Good | Lowering and rolling movements: AUC = 0.83 | + |

Table 2.17 Known groups validity of static balance assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Paediatric Clinical Test of Sensory Interaction in Balance* | | | | | | |
| Gagon et al., 2004 | n = 40<br>n = 40 control | ~12 years | 1 week<br>4 weeks<br>3 months | Adequate | p = 0.46-0.49 | - |
| *Modified Balance Error Scoring System* | | | | | | |
| Howell et al., 2019 | n = 15<br>n = 31 control | ~16 years | ~7 days | Very Good | p = 0.23-0.06 | - |
| VanDeventer et al., 2021 | n = 81<br>n = 58 control | ~14 years | ~6 days | Very Good | p = 0.019-<0.001 | + |
| *Virtual Reality Balance* | | | | | | |
| Teel et al., 2015 | n = 28<br>n = 94 control | >18 years | ~7-10 days | Doubtful | p = 0.006-<0.001 | + |

Table 2.18 Know groups validity of dynamic balance assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Bruininks-Oseretsky Test of Motor Proficiency* | | | | | | |
| Gagon et al., 2004 | n = 40 | ~12 years | 1 week | Adequate | p = 0.001 | + |
| | n = 40 | | 4 weeks | | p = 0.002-0.001 | + |
| | control | | 3 months | | p = 0.024-0.02 | + |
| *Postural Stress Test (PST)* | | | | | | |
| Gagon et al., 2004 | n = 40 | ~12 years | 1 week | Adequate | p = 0.027 | + |
| | n = 40 | | 3 months | | p = 0.024 | + |
| | control | | | | | |

Table 2.19 Known groups validity of gait-based assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Tandem Gait* | | | | | | |
| VanDeventer et al., 2021 | n = 81 | ~14 years | | Very Good | p = <0.001 | + |
| | n = 58 | | | | | |
| | control | | | | | |
| *Dual-Task Tandem Gait* | | | | | | |
| VanDeventer et al., 2021 | n = 81 | ~14 years | | Very Good | p = 0.002-<0.001 | + |
| | n = 58 | | | | | |
| | control | | | | | |
| *Gait* | | | | | | |
| Howell et al., 2019 | n = 15 | ~16 years | ~7 days | Very Good | p = 0.006 | + |
| | n = 31 | | | | | |
| | control | | | | | |
| *Instrumented Gait* | | | | | | |
| Howell et al., 2019 | n = 15 | ~16 years | ~7 days | Very Good | p = 0.46-0.002 | -/+ |
| | n = 31 | | | | | |
| | control | | | | | |
| Wilkerson et al., 2021 | n = 9 | ~25 years | ~3 ± 2.2 years | Adequate | Lateral: p = 0.76-0.003 | -/+ |
| | n = 7 | ~20 years | | | Diagonal: p = 0.96-0.46 | - |
| | control | | | | | |

Table 2.19 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Dual-Task Gait* | | | | | | |
| Howell et al., 2019 | n = 15<br>n = 31<br>control | ~16 years | ~7 days | Very Good | p = 0.29 | - |
| *Instrumented Dual-Task Gait* | | | | | | |
| Howell et al., 2019 | n = 15<br>n = 31<br>control | ~16 years | ~7 days | Very Good | p = 0.68-0.07 | - |

Table 2.20 Known groups validity of task-specific motor assessments

| | Participants | Age | Time since concussion | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Run-Roll-Aim* | | | | | | |
| Prim et al., 2019 | n = 33<br>n = 50<br>control | ~26 years | ~2 weeks | Very Good | p = <0.01 | + |
| *The Portable Warrior Test of Tactile Agility* | | | | | | |
| Cecchini et al., 2021 | n = 64<br>n = 59<br>control | ~29 years<br>~29 years | ~5 months | Very Good | Time to complete<br>Single-task: p = <0.001<br>Dual-task: p = 0.05 | + |
| *Instrumented Portable Warrior Test of Tactile Agility* | | | | | | |
| Favorov et al., 2021 | n = 42<br>n = 57<br>control | ~29 years<br>~29 years | ≤ 2 years | Very Good | Lowering portion:<br>p = <0.0001 | + |

Table 2.21 Measurement error of static balance assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Balance Error Scoring System* | | | | | | |
| Finnoff et al., 2009 | n = 3 | >18 years | | Adequate | MDC test-retest = 7.3<br>MDC inter-tester = 9.4 | ? |

Table 2.21 continued

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Instrumented Balance Error Scoring System* | | | | | | |
| Riemann et al., 1999 | n = 111 | ~20 years | | Adequate | SEM = 0.00-0.45 | ? |
| *Sensory Organisation Test* | | | | | | |
| Register et al., 2013 | n = 38 | ~20 years | | Very Good | SEM = 4.92-5.00 | ? |
| *Instrumented SWAY Balance* | | | | | | |
| Amick et al., 2015 | n = 24 | ~25 years | 1 week | Adequate | SEM = 7.56 | ? |
| | | | 2 weeks | | SEM = 5.82 | ? |
| | | | 3 weeks | | SEM = 5.77 | ? |
| *Instrumented Single Leg Stance* | | | | | | |
| Westwood et al., 2020 | n = 12 | ~22 years | ~10 days | Adequate | SEM = 0.41-2.97 | ? |

Table 2.22 Measurement error of dynamic balance assessments.

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Instrumented Limits of Stability Test* | | | | | | |
| Lininger et al., 2018 | n = 27 | ~24 years | ~7 days | Very Good | SEM = 0.35-1.17 | ? |
| *Dynamic Postural Stability Index (DPSI) and DPSI with Dual-Task* | | | | | | |
| Westwood et al., 2020 | n = 12 | ~22 years | ~10 days | Adequate | DPSI SEM = 0.0047-0.023<br>DPSI+DT SEM = 0.004-0.019 | ? |

Table 2.23 Measurement error of gait-based motor assessments

| | Participants | Age | Time Interval | Method Quality | Results | Measurement Quality Rating |
|---|---|---|---|---|---|---|
| *Tandem Gait* | | | | | | |
| Howell et al., 2019 | n = 32 | ~14 years | | Adequate | RCI = 5.3 sec | ? |
| Howell et al., 2019 | n = 15 | ~16 years | ~7 days | Very Good | SEM = 0.07 | ? |
| *Dual-Task Tandem Gait* | | | | | | |
| Howell et al., 2019 | n = 32 | ~14 years | | Adequate | RCI = 8.5 sec | ? |
| Howell et al., 2019 | n = 15 | ~16 years | ~7 days | Very Good | SEM = 0.06 | ? |
| *Instrumented Dual-Task Tandem Gait* | | | | | | |
| Howell et al., 2020 | n = 20 | ~22 years | ~2 weeks | Adequate | SEM = -0.08-1.35 | ? |

Table 2.23 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| *Instrumented Gait* | | | | | | |
| Howell et al., 2016 | n = 45 | ~15-22 years | Initial<br>1 week<br>2 weeks<br>1 month<br>2 months | Very Good | MDC = 0.88-7.42 | ? |
| Kuznetsov et al., 2018 | n = 32 | ~25 years | ~7 days | Very Good | Temporal metrics SEM = 0.02-0.34<br>Spatial metrics SEM = 0.32-4.10 | ? |
| Howell et al., 2020 | n = 20 | ~22 years | ~2 weeks | Adequate | Discrete trials SE = -0.50-0.04<br>2 min trial SE = -1.32—0.08 | ? |
| Howell et al., 2019 | n = 31 | ~16 years | ~7 days | Very Good | SEM = 0.07-0.08 | ? |
| *Instrumented Dual-Task Gait* | | | | | | |
| Howell et al., 2016 | n = 45 | ~15-22 years | Initial<br>1 week<br>2 weeks<br>1 month<br>2 months | Very Good | MDC = 0.84-6.76 | ? |
| *Battery Gait Assessment* | | | | | | |
| Howell et al., 2019 | n = 15 | ~16 years | ~7 days | Very Good | SEM = 0.05 | ? |

#### 2.3.3.8 Known group validity

For known group validity, static balance included paediatric clinical test of sensory interaction in balance (PCTSIB) (n = 1), mBESS (n = 2), virtual reality balance (n = 1). Outcome variables time and number of errors for non-instrumented assessments. Instrumented versions assessed COP displacement. Time since concussion averaged 7-days for all assessments. Dynamic assessments included Bruininks-Oseretsky test of motor proficiency (n = 1), and Postural Stress Test (PST) (n = 1). Outcome measures for PST assessed weight required for counterbalance. Bruininks-Oseretsky test of motor proficiency measured number of errors and time to complete. Both assessments were conducted at 1-week and 3-month time periods. Gait assessments included tandem gait (n = 1), dual-task tandem gait (n = 1), gait (n = 1), instrumented gait (n = 1), dual-task gait (n = 1), instrumented dual-task gait (n = 1). Time since concussion ranged from 7-days to 3-years. Other assessments included the run-roll-aim task (n = 1) and the Portable Warrior Test of Tactile Agility (n = 2). Both mBESS and virtual reality static balance showed significant between group differences when conducted within 10-days of sustaining a concussion. Both dynamic assessments displayed significant between group differences up to three months post-concussion. However, reliance on specialised equipment reduces their feasibility for clinicians. Gait assessments including single and dual-task tandem gait, and gait also reported significant between group differences when conducted 7-days post-concussion.

Athletes from contact and non-contact sports (n = 2,533; 97%) were included, as well as military personnel (n = 83; 3%) who had been diagnosed with concussion. The most common test was the mBESS, representing 16% of all tests.

#### 2.3.3.9 Measurement error

The measurement error of 13 lower limb motor assessments was assessed over 10 different studies. Quality ranged from adequate to very good. Assessments were categorised into static balance (n = 5), dynamic balance (n = 2), and gait (n = 6). Static balance assessments included BESS (n = 1), instrumented BESS (n = 1), SOT (n = 1), instrumented SWAY balance (n = 1), instrumented single leg stance (n = 1). Studies reported the standard error of the measure (SEM), limits of agreement (LOA), or minimal detectable change (MDC). The instrumented

BESS (SEM = 0.04 – 0.45) and instrumented single leg stance (SEM = 0.49 – 2.97) displayed the lowest SEM (202, 224). Dynamic assessments included the instrumented limits of stability test (n = 1) and the DPSI (n = 1). Both single-task (SEM = 0.0047 – 0.023) and dual-task (SEM = 0.004 – 0.019) variations displayed the lowest SEM (224). Gait assessments included tandem gait (n = 2), dual-task tandem gait (n = 2), instrumented dual-task tandem gait (n = 1), instrumented gait (n = 4), instrumented dual-task gait (n = 1), and a gait battery assessment (n = 1). All gait assessments displayed low SEM across trials, therefore promoting the use of instrumented or non-instrumented gait assessments as acceptable tools to measure motor changes. A summary of study characteristics is presented in Table 2.2. Full details of the studies' characteristics are presented in Supplementary Table 2.21 through to Table 2.23.

## 2.4 Discussion

This systematic review aimed to (1) consolidate the reliability and validity of motor function assessments across the time course of concussion management, and (2) summarise their feasibility for clinicians and other end-users. In general, instrumented assessments providing objective analysis tended to offer superior reliability and validity compared with non-instrumented, subjective assessments, however, may not be feasible for all users. Gait-based assessments showed the best reliability, with instrumented methods offering a range of outcome variables. Sensitivity is improved with an objective method of assessing performance, on more complex tasks, and during the acute stages of injury. Non-instrumented assessments offer greater practical utility, but this may come at the expense of reliability and validity, particularly beyond 2-weeks post-concussion. Overall, each assessment had limitations and practitioners should be mindful of these when selecting the most appropriate assessment for their setting. However, best practice encourages practitioners to use a variety of assessments within a battery to accurately assess the multitude of symptoms experienced. Solely relying on single-assessor, subjective diagnostic test to guide the RTP or return-to-duty process should be avoided. When selecting appropriate assessments and interpreting results, reliability, validity, and feasibility should be considered. Where possible, practitioners should look to use instrumented assessments for which the error, reliability and validity has been established, and a range of outcome variables can be monitored.

### 2.4.1 Reliability

In general, objective testing from instrumented assessments offered greater test-retest when compared with subjective. Instrumented assessments also offer clinicians more detailed measures of motor function, thus providing a more comprehensive analysis of readiness for RTP (12).

2.4.1.1 Static balance assessments

Test-retest reliability for static assessments varied between subjective and objective measurements. In general, non-instrumented assessments relying on subjective interpretation, such as the BESS and mBESS displayed *insufficient* reliability across multiple testing points ranging from 2-days to 20-months (198, 199, 213, 215, 216). However, improved reliability was reported for both of these assessments when an increased number of trials was performed and a minimum of two assessors were present (198). Due to a suggested learning effect associated with the BESS and mBESS, it was found that allowing a practice trial followed by 2-3 subsequent test trials produced the best reliability, taking around 10 minutes to administer (198). The BESS displayed greatest test-retest reliability when more than 2 trials of 4 conditions (excluding double leg stance) were performed (199). This differs from standard practice, where practitioners are to perform a single assessment as a means of evaluating balance deficits. Although this approach is more feasible for clinicians, only one study displayed *sufficient* reliability with one trial ($r = 0.78$) (197), with other studies showing greater reliability with multiple trials (198, 199). Best practice would be to perform multiple trials as a single trial likely jeopardises the reliability of the assessment, limiting its justification for inclusion. Therefore, clinicians need to decide which takes priority; reliability of the measure, or practicality of its implementation. Differences in interpretation of errors between assessors also contributes to the *insufficient* reliability of these tools (255). These differences between assessors may be exacerbated when performed on concussed individuals during acute stage of injury due to an increased number of balance errors offering a greater bandwidth for disagreement to occur. Previous findings have shown that making recommendations based on the average of 3 different clinicians' assessments and providing clear guidelines on how to administer and score the test may assist in improving reliability (255), although this may not be viable in many practical settings. Instrumented static balance assessments that offered objective outcomes displayed *sufficient* reliability, with the instrumented BESS, balance tracking system, and BAM superior to other instrumented static assessments. Of these the BAM, utilising accelerometers, may be a more feasible and cost-effective option for clinicians as opposed to using force plates. Being aware of the inherent noise and the MDC of these assessments is vital for making decisions on changes in performance. For example, the BESS has shown MDC of 7.3 errors for test-retest (202), however studies have shown an average of 3-7 errors are typically performed post-concussion (73, 256). Therefore, the test may lack the sensitivity to detect important balance deficits beyond the acute stages of injury. Instrumented static assessments (i.e., with a force plate or IMU) should be selected over non-instrumented methods wherever possible. If practitioners are working in settings that only permit non-instrumented, static assessments, they should ensure that there is *sufficient* familiarisation prior to scoring, use multiple assessors, and there are clear scoring guidelines. If these criteria cannot be met, justification for conducting the assessment beyond diagnosis should be scrutinised in future standardised assessment protocols.

#### 2.4.1.2 Dynamic balance assessments

Few studies analysed the reliability of dynamic assessments, with results favouring the use of dynamic assessments over static. Only one study assessed the reliability of a non-instrumented dynamic motor response assessment with clinical reaction time (modified drop-stick test) (213). While this study demonstrated *insufficient* test-retest reliability (ICC = 0.32) over an 11-month timeframe (213), reliability may be improved over shorter time periods. Instrumented dynamic assessments, on average, displayed clinically acceptable reliability (r = 0.32 to 0.99) when conducted within 10-days. Force plates sampling at 100 – 1200 Hz were shown to be useful when assessing postural sway (201, 224), however may not be readily available for all clinicians. Alternatively, IMUs also demonstrated *sufficient* reliability during the Y balance test (ICC = 0.76 – 0.99) (200), and may be a more feasible option for clinicians. For those who do not have access to the required equipment, non-instrumented gait assessments are recommended.

#### 2.4.1.3 Gait assessments

In general, gait assessments were seen to have the greatest test-retest reliability when compared to static and dynamic balance tests. Non-instrumented tandem gait assessments focusing on temporal gait parameters (i.e., time to complete, cadence) showed *sufficient* reliability across most studies (220, 221, 242-244). However, test-retest reliability was *insufficient* when conducted beyond 2-months. This presents an issue when relying on pre-season baseline testing of tandem gait (such as during the SCAT6 protocol (134)) to interpret post-concussion scores. Therefore, if subjective assessments are to be used, it is recommended that practitioners are aware of the reliability and conduct baseline assessments in line with these timepoints.

Instrumented gait assessments assessing temporal and spatial (i.e., stride length) gait parameters also demonstrated *sufficient* reliability. Lumbar and foot mounted IMUs were clinically acceptable and offer clinicians an inexpensive and reliable alternative to laboratory equipment (246-248). Smartphone apps measuring movement vectors also displayed *sufficient* test-retest reliability when firmly positioned on the body (246, 248, 249, 251), but exhibited *insufficient* reliability when held in the hand. Measures of step length, step time, gait velocity, and cadence when derived from placement at the lumbar spine, or pelvis (anteriorly via belt) were most reliable (249). The use of laboratory equipment such as 3D motion capture or a GAITRite system also displayed *sufficient* reliability across trials (249, 250), however the associated equipment costs and expertise requirements reduce the feasibility of these tools in most situations. Feasibility is also compromised due to the difficulty in obtaining baseline pre-injury scores meaning normative or control comparisons are needed. Researchers should look to develop a more readily available means of capturing pre-concussion baseline scores using commercially available technologies such as smartphones, IMU and global navigation satellite systems (GNSS) devices.

#### 2.4.1.4 Considerations

Clinicians should be encouraged to implement dynamic balance or gait-based assessments as a part of a comprehensive and multifaceted concussion assessment approach, due to their higher test-retest reliability than static approaches. As previously mentioned, consistency across trials allows variations in motor strategies to be more easily detected, when a concussion is sustained (180). Multiple trials, with the average taken, should be completed if performing non-instrumented static assessments (198), with the assessment made by multiple clinicians, in preference to one to minimise noise in the measurement and allow for smaller changes in performance to be detected as real changes (255). Additionally, clinicians should also be mindful of time between repeated measures. Objective measures drawn from instrumented assessments provide better test-retest reliability, places less pressure on the clinician and limits the ability of players to hide symptoms. The use of more clinically practical tools such as IMUs or smartphones, which are reliable for use in dynamic and gait-based tasks (200, 246-249), should be encouraged.

### 2.4.2 Validity

Validity ratings of assessments ranged from *sufficient* to *insufficient* based on COSMIN guidelines (194). In general, dynamic balance and gait assessments offered greater validity when compared with static assessments. However, validity was compromised across all assessments as time since concussion increased beyond seven days, which is likely an artefact of partial or complete recovery from the concussion beyond this point.

#### 2.4.2.1 Static balance assessments

Construct validity for static assessments varied, with instrumented assessments offering better validity when compared with non-instrumented. Commonly used subjective assessments; BESS and mBESS displayed *insufficient* ability to discriminate between groups when performed more than 48-hours post-concussion, however had *sufficient* validity when performed within 24-hours. (114, 156, 207, 208, 217-219). Therefore, these assessments may aid in diagnosis, however caution should be applied if implementing as part of a RTP protocol. Traditional models of SRC management include the assessment of subjective static balance (mBESS) to assist with decisions regarding RTP (12, 187). Whilst instrumenting these assessments with a force plate or IMU improves sensitivity, they are still limited beyond two weeks post-injury (146, 226). Motor function entails a complex hierarchy of integration between systems, and therefore needs to be assessed along a spectrum of varying complexity (12). During the acute stages, athletes demonstrate a significant increase in errors when performing the mBESS, however return to baseline 3-5 days post-concussion (12, 257). Due to the gross outcome measures, and suggested learning effect, it is believed that these assessments are unable to challenge the sensorimotor system to identify any underlying deficits in motor function (12). Further, these simple static tasks are not reflective of the complex dynamic athletic tasks performed, such as running and tackling.

Virtual reality static balance using a 3D projection system displayed *sufficient* ability to discriminate between concussed and non-concussed (0.857) when conducted 10-days post-concussion (210). This highlights the promise for the use of virtual reality technology in monitoring concussion symptoms, although the equipment is not readily available in most practical settings, therefore reducing its feasibility.

2.4.2.2 Dynamic balance assessments

In general, dynamic balance assessments displayed better construct validity than static balance assessments. However, these were still limited beyond 2-weeks post-concussion. Findings highlighted the importance of test selection relative to the population being assessed. In particular, the KPR test displayed *sufficient* sensitivity for children, and may be a feasible option for assessing readiness for RTP in this population (211). The PANESS, community balance and mobility scale, and instrumented Y balance test all demonstrated *sufficient* sensitivity on adult populations (0.76 – 1.00) when conducted within 2-weeks post-concussion (235, 236, 240, 241). Like static assessments, these tasks are unlikely to challenge the neuromuscular system beyond the acute stage of injury. Using them to monitor changes across a graduated RTP protocol may not be best practice, particularly in concussions where symptoms persist beyond 2-weeks.

2.4.2.3 Gait assessments

Validity of gait assessments varied amongst studies. The functional gait assessment ranged from *insufficient* to *sufficient* sensitivity (0.05 – 0.75) (240, 241), with higher sensitivity found when performing the assessment within 1-week post-concussion. Therefore, clinicians should be cautious if implementing this assessment tool beyond this time. Assessment of gait speed during normal and tandem gait, in general, demonstrated *sufficient* sensitivity when conducted within 1-week post-concussion (114, 151, 156, 217, 240, 241, 244).Dual-task gait displayed sufficient sensitivity for children when conducted within 2-weeks of sustaining a SRC (245). However, clinicians should be mindful of using gross measures of gait (e.g., time taken), due to the lack of outcome measures provided. The addition of a cognitive task (dual-task) improved sensitivity for most studies (114, 151, 244) when completed 1-week post-concussion (218). Instrumented gait assessments had mixed results. Assessment of single- and dual-task gait using lumbar and foot mounted IMUs amongst adult populations within five days of sustaining a concussion, demonstrated *insufficient* sensitivity for gait speed, cadence, and stride length when comparing to normative reference values (115). However, measures of single-task gait velocity and cadence using a smartphone affixed to the lumbar spine demonstrated *sufficient* AUC and between group differences for adolescent populations with concussion when conducted 1-week post-injury. (0.76-0.79) (151). Like tandem gait assessments, the addition of a cognitive task improved sensitivity. Dual-tasks conditions aim to highlight potential deficits in attention allocation and executive function. Typically, these are observable through increased errors in a cognitive task, or variability in gait tasks (258). Although these assessments tend to provide greater sensitivity than single-task versions, limitations still exist beyond 2-weeks post-concussion (253). The use of a virtual reality system 3-months post-concussion displayed *sufficient* AUC (0.79-0.84) and significant between group differences for reaction time and lateral movement asymmetries during a reactive movement task (252). However, further research is warranted due to the small sample size used within this study. Additionally, the need for normative data currently reduces the utility of this assessment. An instrumented battery gait assessment conducted 1-week post-concussion, consisting of; gait velocity, cadence, tandem gait time, and dual-task tandem gait time displayed *sufficient* sensitivity and specificity when all measures were combined (AUC = 0.91) (151). However, time taken to conduct may be a barrier. Clinicians are encouraged to implement gait assessments where possible due to their ability to better classify those with and without SRC. Instrumented versions using laboratory equipment or more feasible tools such as IMUs or smartphones are the preferred option.

#### 2.4.2.4 Other assessments

The military specific run-roll-aim assessment demonstrated statistically significant differences between concussed and control participants for ability to complete the task within 2-weeks post-concussion (206). No differences were found between total time, number of correct targets identified, or delay in reaction time for cognitive stimulus, otherwise referred to as Stroop effects committed. The Portable Warrior Test of Tactile Agility demonstrated statistically significant differences in time to complete for both single-and dual-task variations (254). The instrumental version of this assessment, utilising IMUs demonstrated *sufficient* ability to discriminate between concussed and control during the 'lowering and rolling' movements (AUC = 0.83) (212). No statistically significant differences for other portions of the assessment were seen.

#### 2.4.2.5 Considerations

In general, instrumented assessments demonstrated a better ability to discriminate between concussed and non-concussed individuals. Measures of static balance were more accurate via the use of force plates (145, 146) or a 3D virtual reality projection system (210). However, limitations surrounding suggested learning effects, and the utility of these devices, such as costs and low ecological validity, does limit their application throughout the management process following concussion. Both instrumented and non-instrumented dynamic balance assessments displayed *sufficient* sensitivity when conducted within 2-weeks post-concussion, therefore offering cost effective and more objective options for clinicians. Assessing time to complete on dual-task tandem gait was shown to be a sensitive and cost-effective assessment that clinicians could easily implement if access to instrumented versions is not feasible (114, 151, 244). However, this does not provide clinicians with a variety of outcome measures, nor does it have any use beyond the acute stages of concussion (1-week). The use of IMUs (lumbar and foot mounted), stop-watches, and smartphones (lumbar mounted) provide improved reliability and validity, however this is surpassed with the use of laboratory equipment (3D motion analysis, GAITRite). Nevertheless, laboratory equipment, although highly accurate, is largely impractical for clinician use due to associated costs, expertise, and requirement of specialised facilities (160).

In general, sensitivity of assessments reduced as time from initial injury increased, which is unsurprising given the varied time course of recovery between individuals. Furthermore, sensitivity of both static (146, 233) and gait (253) assessments was reduced beyond 2-weeks post-concussion, therefore clinicians must be cautious when using these assessments as a RTP measure beyond this timeframe. Athletes returning to play following a concussion have shown an increased risk of acute musculoskeletal injury (100, 101, 259). It is suggested that subclinical neuromuscular deficits may linger beyond expected recovery timeframes, however due to poor assessment availability and limited research surrounding best-care concussion management, many of these changes go undetected (12, 100, 101, 259). This review provides clinicians with reliability and validity measures of assessments to allow a more educated selection of tests. However, it also highlights the problems with concussion management protocols, specifically the over-reliance on tools not initially designed to inform RTP decisions.

## 2.5 Feasibility and utility

This review aimed to summarise the reliability and validity of lower limb motor assessments for the management of SRC. However, what should not be overlooked, is the clinical utility and feasibility of such assessments and their seamless integration within a RTP or return-to-duty protocol. Aside from the reliability and validity of a measure, stakeholders must also consider other factors such as interpretability of outcomes, cost of equipment, expertise required, and time needed for implementation and analysis of results, when developing assessment protocols. In general, instrumented assessments demonstrated better test-retest reliability across multiple time periods as well as better ability to discriminate between concussed and non-concussed individuals. Of these, laboratory assessments using force plates, 3D motion capture, or pressure sensitive walkways provided clinicians with more accurate objective measures. However, these display low ecological validity for the assessment of field-based sports due to the controlled environmental conditions (160) and lack of flexibility in tasks that can be performed, and therefore may have poor crossover to the stochastic nature of sports competition. Equipment and facility requirements are typically associated with high cost, and therefore not feasible for most team sports (160). Furthermore, the need of trained personnel to collect and analyse the data may act as further barriers to their uptake within practice.

Other tools used for instrumented assessments included IMUs and smartphone devices. These tools were shown to have better test-retest reliability and validity for most assessment categories (static, dynamic, gait). Studies included in this review assessed validity and reliability of lumbar and foot-mounted IMUs (200, 246-248). Test-retest reliability for dynamic and gait assessments using these devices were similar to those from laboratory assessments. Similar findings were associated with the use of smartphone devices, displaying *sufficient* test-retest reliability during gait assessments (231, 249). Although they achieved poorer validity than laboratory equipment, IMUs and smartphone devices offered clinically acceptable validity, specifically during dynamic balance and gait assessments (115, 151, 236, 239, 253). In regard to interpretability of results, cadence and gait velocity metrics derived from IMUs and smartphones displayed sufficient ability to discriminate between concussed and non-concussed. Typically, these measures are made readily available for clinicians when using the appropriate software for the respective device and therefore avoids the need for additional analysis. As such, the lower cost, autonomy for analysis, and greater portability of these devices may improve their uptake in the field. These devices may offer practitioners the ability to identify at-risk individuals who require further investigation through more in-depth assessments. Efforts should be made to make these instrumented assessments more feasible for end-users without compromising reliability or validity. Utilising technologies such as IMUs embedded in current wearable technologies (e.g., GPS units, smartphones and watches) should be explored further.

## 2.6 Conclusions

Based on the findings from this review, clinicians are encouraged to implement instrumented or non-instrumented dynamic balance or gait assessments as part of a battery of assessments and not in isolation. Instrumented assessments utilising more complex gait tasks should be encouraged to add resolution to existing RTP protocols. On average, static assessments displayed *insufficient* test-retest reliability and validity for the management of SRC. If practitioners do not have the resources to perform instrumented tests, it is recommended that they consider the reliability and validity issues that potentially limit the simpler test options. Practitioners may refer to Table 2.24 proving a summary of the assessments included within this review, highlighting the pros and cons alongside the key takeaways. Future research should look to establish standardised protocols and best practice for monitoring motor function during the RTP period and beyond. Developing the use of accessible technologies such as IMUs, smartphones and the use of marker-less tracking to monitor gait function is an important step for concussion management. Furthermore, understanding how movement changes under more context specific scenarios, where fatigue, decision-making, and the performance of more complex movements occur, is warranted.

Table 2.24 Summary and key points of assessment categories included within the review

| Assessments | Pros | Cons | Key takeaway |
|---|---|---|---|
| *Non-instrumented static balance assessments* | | | |
| Balance Error Scoring System (BESS)<br><br>Modified BESS (mBESS)<br><br>Single leg stance<br><br>Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)<br><br>Paediatric Clinical Test of Sensory Interaction in Balance (PCTSIB) | Easy and quick to administer<br><br>Minimal equipment required<br><br>Provide general insight into motor control during simple, static tasks | Subjective interpretation of the number of errors<br><br>Poor inter-rater reliability when no clear assessment criteria are provided<br><br>Poor discriminant ability beyond acute stages of concussion (7-days post-concussion) | Effective assessments to assess deficits in static balance immediately following a concussive event and for the acute stages of concussion management (e.g., 7-days post-concussion)<br><br>Not reflective of the activities athletes will be required to perform upon RTP |
| *Instrumented static balance assessments* | | | |
| Instrumented BESS<br><br>Instrumented mBESS<br><br>Instrumented single leg stance<br><br>Balance Accelerometry Measure (BAM)<br><br>Balance tracking system<br><br>SWAY Balance<br><br>Sensory Organisation Test (SOT)<br><br>Physiological vibration acceleration (Phybrata) System<br><br>Instrumented Modified Clinical Test of Sensory Interaction in Balance (MCTSIB)<br><br>Virtual Reality Balance | Most assessments that do not require specialised equipment are quick and easy to administer<br><br>Assessments utilising IMUs provide an affordable and accessible way to gain objective data from traditionally subjective assessments | Some assessments (e.g., SOT, balance tracking system, MCTSIB, virtual reality balance) require specialised laboratory equipment (e.g. specialised system, force plates) that may be costly and/or unattainable for most team-sport clinicians | Effective assessments to assess deficits in static balance immediately following a concussive event and for the acute stages of concussion management (e.g., 7-days post-concussion)<br><br>Not reflective of the activities athletes will be required to perform upon RTP<br><br>Where possible, instrumented static balance assessments should be used over non-instrumented static balance assessments |

Table 2.24 continued

| | | | |
|---|---|---|---|
| *Non-instrumented dynamic balance assessments* | | | |
| Clinical reaction time<br><br>Kash Pulse Recovery (KPR)<br><br>Physical and Neurological Examination of Subtle Signs (PANESS) | Clinical reaction time and KPR assessments are quick and easy to administer<br><br>Can provide more comprehensive assessment of motor function when compared to static balance assessments | The time required to conduct the PANESS assessment may limit the uptake amongst clinicians with strict time constraints | Effective assessments to assess deficits in dynamic balance immediately following a concussive event and for the acute stages of concussion management (e.g., 7-days post-concussion)<br><br>Not reflective of the activities athletes will be required to perform upon RTP<br><br>Where possible, non-instrumented dynamic balance assessments should be used over or in conjunction with instrumented and non-instrumented static balance assessments |
| *Instrumented dynamic balance assessments* | | | |
| Instrumented Y Balance Test (YBT)<br><br>Instrumented Limits of Stability Test (Instrumented LOS)<br><br>Dynamic Postural Stability Index (DPSI) and DPSI with Dual-Task<br><br>Postural Stress Test (PST)<br><br>Computer-Assisted Rehabilitation ENvironment (CAREN) System<br><br>Battery combined: Grip strength, single leg jump, MCTSIB<br><br>Bruininks-Oseretsky Test of Motor Proficiency | Assessments utilising IMUs provide an affordable and accessible way to gain objective data from traditionally subjective assessments<br><br>Instrumented dynamic balance assessments may provide a more comprehensive insight into motor strategies following concussion | Most assessments (e.g., DPSI, CAREN system, battery assessment, LOS, Bruininks-Oseretsky Test of Motor Proficiency, PST) require specialised laboratory equipment (e.g. specialised system, force plates) that may be costly and/or unattainable for most team-sport clinicians<br><br>The time required to run some assessments (e.g., battery assessment) may limit their feasibility amongst clinicians with strict time constraints<br><br>Limited discriminant ability beyond acute stages (e.g., 7-days) following concussion | Effective assessments to assess deficits in dynamic balance immediately following a concussive event and for the acute stages of concussion management (e.g., 7-days post-concussion)<br><br>The instrumented YBT utilising IMUs offers an affordable and easy to administer assessment for clinicians to objectively assess dynamic balance<br><br>Where possible, instrumented dynamic assessments should be used over, or in conjunction with, non-instrumented dynamic balance assessments |

Table 2.24 continued

| | | | |
|---|---|---|---|
| *Non-instrumented gait assessments* | | | |
| Community Balance and Mobility Scale<br><br>Timed Up and Go (TUG)<br><br>Walking on Balance Beam<br><br>Tandem Gait<br><br>Dual-Task Tandem Gait<br><br>Complex Tandem Gait<br><br>Gait<br><br>Dual-Task Gait | Most assessments are easy and quick to administer<br><br>Assessment of gait is more relevant to sport demands following RTP<br><br>Dual-task gait assessments may provide more insight into neurological changes post-concussion | Time to complete and subjective observation of single-task gait strategies are the only measures of performance for these assessments. Time to complete may be too blunt of a measure to provide insight into any subtle changes in gait mechanics | The assessment of gait may be more applicable to RTP requirements for athletes following concussion. However, clinicians should be aware of the limitation associated with time to complete a task<br><br>Dual-task gait assessments may provide a more comprehensive insight into gait strategies and cognitive processing than single-task gait<br><br>Where possible, gait assessments should be used over or in conjunction with instrumented or non-instrumented dynamic balance assessments |
| *Instrumented gait assessments* | | | |
| Battery combined: Berg balance, Balance Evaluation Systems Test (BESTest), Dynamic gait index, High-Level Mobility Assessment Tool (HiMat)<br><br>Battery Combined: gait, tandem gait, dual-task tandem gait<br><br>Instrumented Dual-Task Tandem Gait<br><br>Instrumented Gait<br><br>Instrumented Dual-Task Gait<br><br>Functional Gait Assessment | Assessments utilising IMUs provide an affordable and accessible way to gain objective data for both single and dual-task gait<br><br>Assessment of gait is more relevant to sport demands following RTP<br><br>Dual-task gait provides additional insight into cognitive processing and the interaction between executive functioning and neuromotor function | Assessments that require specialised equipment (e.g., motion capture, specialised walking force plates) that may be costly and/or unattainable for most team-sport clinicians<br><br>Some assessments (e.g., battery assessments) may be too time consuming and therefore have limited uptake amongst clinicians<br><br>Data is limited to walking gait and therefore conclusions for higher velocities cannot be made | The assessment of instrumented gait may be more applicable to RTP requirements for athletes following concussion. In addition, data obtained from wearable devices, such as IMUs, can provide insight into variations in gait strategies beyond more blunt measures of time to complete.<br><br>Where possible, instrumented gait assessments should be used over or in conjunction with non-instrumented gait assessments<br><br>Future research should look to assess gait strategies at greater velocities |

Table 2.24 continued

| | | | |
|---|---|---|---|
| *Non-instrumented task specific assessments* | | | |
| Run-Roll-Aim<br><br>Portable Warrior Test of Tactile Agility | Good assessments for individuals involved in combat specific sports | Limited application to most team-sport athletes | Good assessment for individuals involved in combat specific sports; has limited application for most team-sport athletes |
| *Instrumented task specific assessments* | | | |
| Instrumented Portable Warrior Test of Tactile Agility | Good objective assessment for individuals involved in combat specific sports | Assessment may require specialised equipment and can be time consuming. This may limit its uptake | Good assessment for individuals involved in combat specific sports; has limited application for most team-sport athletes |

## 2.7 Linking paragraph

Findings of the systematic review highlight the inadequacies of the current management processes for sports-related concussions. Current clinical motor assessments are heavily reliant on subjective interpretation, seem to be incapable of detecting subtle variation in motor strategies, and display poor sensitivity beyond acute stages of concussion. These attributes limit their use as tools to assess RTP readiness. Alternatively, laboratory assessments provide high-quality objective data and are better indicators of subtle motor strategy variations. However, the cost and expertise requirements limit their feasibility for everyday use by most practitioners. The use of wearable IMUs has shown promise for detecting motor strategy changes following concussion and offer a more cost-effective and portable alternative for practitioners. Despite this, their uptake in practice is limited which may be due to their use within concussion management, being isolated to clinical, controlled environments via either balance or walking assessments. This either means practitioners need access to sites, equipment and expertise that may not be readily available. To overcome some of these barriers, assessing the reliability and sensitivity of wearable IMU devices in an outdoor setting, real-world environment is warranted. This research would provide practitioners with more informative data, potentially increasing their uptake within the field, promoting a safer return to sport and more regular feedback to inform recovery in response to various athlete rehabilitation strategies.

# Chapter 3

# Extended Methods

## 3.1 Overview

The association between sports-related concussion (SRC) and subsequent injury following return to play (RTP) has been documented (3-6). An increase in lower limb musculoskeletal injury incidence has been observed amongst those returning from concussion (3-6). This is suggestive of potential subclinical issues in motor control that are not detected during standard management. Specifically, findings of the systematic review have questioned the sensitivity of currently implemented assessments, particularly beyond the acute (<7 days) stages of concussion. Current protocols for the assessment of SRC promote the use of a multifaceted approach, allowing assessment of cognitive, physical, and psychological symptoms. Despite this, commonly used subjective assessments lack the requisite objectivity and have low sensitivity, meaning subclinical changes in function may go undetected (207, 208, 213) and those recovering from concussion may experience an increased rate of injury following RTP. Laboratory assessments appear to be superior diagnostic tools in assessing concussion symptoms (99, 146). However, the specialist equipment, associated costs, and expertise requirements limit their feasibility for most team sport practitioners. Emerging evidence has supported the use of portable microtechnology, such as inertial measurement units (IMUs), as a more feasible alternative to detect changes in motor strategies following concussion (200). Such devices have demonstrated sufficient discriminative ability between concussed and non-concussed populations during clinical gait-based tasks (144, 145, 168). No studies have assessed the use of (1) thoracic mounted devices for (2) gait tasks performed on grassed surfaces at (3) velocities other than walking pace.

The studies included in this thesis aimed to investigate the use of thoracic, lumbar, and ankle mounted IMUs and their potential to allow continuous measurement of recovery following concussion in a team sport environment. This chapter provides an extended outline of the methodology used to achieve these aims. Specific attention is given to the study protocol, its implementation, the outcome measures, and the validity and reliability of instruments used.

## 3.2 Participants

Males and females aged between 15 to 35 years were eligible to participate in this project. Participant age criteria was determined based on several reasons. Firstly, this thesis aimed to capture changes in motor performance across youth and adult sporting communities. Therefore, it was decided to include minors and adults. Secondly, the Australian Sports Commission data outlined that youth (15 to 17 years), and young adult (18 to 35 years) populations had the highest participation in organised sports when compared to adults over 35 years (260). Finally, the average age for senior Australian football (AFL) players and rugby union players typically ranges from 24 to 27 years (261, 262). Therefore, the age criteria requirements ensured findings from this project were applicable to professional level competitions.

Participants were also free from any neurological disorders and had no lower limb injuries within three months of testing. Average RTP time frames from lower-limb injuries (knee, ankle, and hamstring) range from 5.5 weeks to 12.7 weeks (263). Therefore, a minimum recovery of three months was deemed necessary to ensure athlete gait was not affected by previously sustained musculoskeletal injury/s. In addition, control participants had no diagnosed concussions within 12-months of testing. Research suggests symptoms of concussion may still be present up to 12-months post-injury (6, 73, 99).

Concussed participants and age, sex, and sport matched controls were recruited via advertisement through local (Queensland, Australia) rugby league and Australian rules football clubs. A larger pool of healthy active controls was recruited via advertisement at Australian Catholic University, Brisbane. Advertisement included posters, emails, and face-to-face discussions. Specific details for the study populations, including participant demographics are provided in chapters four and five.

### 3.3 Protocol

Prior to testing, concussed participants were required to complete the post-concussion symptom scale (PCSS), which included 22-items that were each self-scored on a 7-point Likert scale (0 to 6). For this assessment, a score of 0 represented no symptoms, while a score of 6 represented severe symptoms (264). Participants were then fitted with a thoracic-mounted wearable device (Apex Pro Series, STATSports) which housed a triaxial accelerometer (100 Hz; ± 16 g), GNSS chip (10 Hz), magnetometer (10 Hz; ± 16 g), and gyroscope (100 Hz; ± 2000 $deg \cdot s^{-1}$) worn in an upright position in a tight-fitting manufacturer provided vest. The device was switched on via a button on the front of the unit approximately 5-minutes prior to testing, ensuring a GNSS signal had been acquired, and remained untouched until the end of the testing session. The GNSS devices used have been shown to offer suitable reliability and validity for measuring instantaneous velocity during straight-line movements (170, 265). Accelerations in the antero-posterior (Z), medio-lateral (X), and vertical (Y) axes were recorded. Participants used the same device for each testing session. Participants were also fitted with three Vicon Blue Trident IMU devices with a triaxial accelerometer (± 16 g) sampling at 1125 Hz (version 10.0.1, VICON, Oxford, UK). One fitted to the medial aspect of each ankle via a manufacturer provided strap, and one fixed to the skin over the approximate height of the 5$^{th}$ lumbar vertebra using double-sided tape and strapping tape (see Figure 3.1). The X, Y, and Z axes for the lumbar mounted devices represented vertical, medio-lateral, and antero-posterior directions, respectively. Due to placement of the ankle mounted units, orientation of the X, Y, and Z axes shifted to antero-posterior, vertical, and medio-lateral, respectively. All devices were fitted and calibrated in line with the manufacturer guidelines prior to each testing session. IMUs were rotated in a figure-eight pattern to calibrate sensors prior to being fitted on participants. Units were connected to an iPad with the Capture U app via Bluetooth. A session was created for both set- and self-paced conditions.

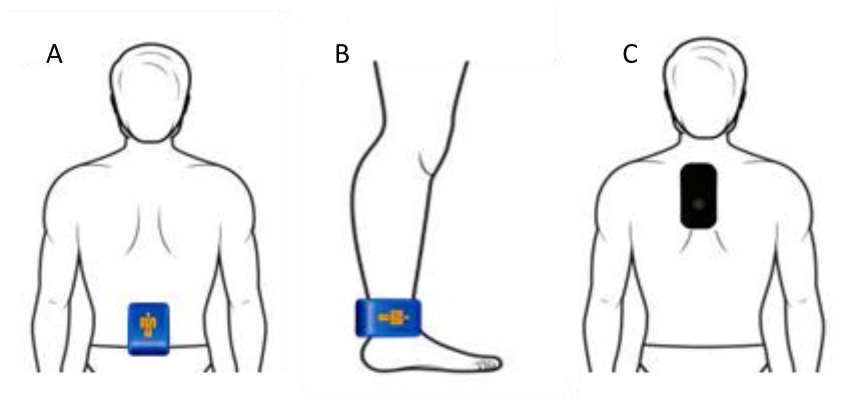


Figure 3.1 Positioning for lumbar (A), ankle (B), and thoracic (C) mounted IMUs. Lumbar-mounted IMUs were attached on the skin at the approximate location of the 5$^{th}$ lumbar vertebra. Ankle-mounted IMUs were attached on the lateral aspect of each ankle via the manufacturer's strap. Thoracic-mounted IMUs were positioned in a manufacturer provided bib over the thoracic spine between the scapulae.

Prior to commencing the protocol, participants completed a standardised warm-up consisting of 10-m jog, 10 × squats, lunges, and hamstring sweeps. Participants were then instructed to complete three × walks, jogs, and runs under set- or self-paced conditions over a 40-m distance before performing the second pacing condition. Whilst the order of activities was the same across sessions, trials were counterbalanced and randomised to complete either the set- or self-paced condition first. Participants were instructed to complete each activity from a static starting position. Rest times between repeated activities and sets were self-selected, but long enough to allow sufficient recovery. To encourage natural gait, all participants wore their normal training footwear and attire which was consistent across each session for each participant.

The protocol was conducted on a flat grass surface away from any foreseeable obstacles or hazards. Four coloured cones (2 blue, 2 white) were used to mark the 40-m straight line. White cones were placed at the start (0-m) and finish (40-m) points of the testing length. The blue cones were then placed at the 10-m and 30-m distances (Figure 3.2). Participants were instructed to complete each activity (walk, jog, run) between the two white cones, accelerating from white to blue, holding speed between the two blue cones, and then decelerating to a standstill by the final white cone. These instructions were provided to each participant at the beginning of each testing session.

During the self-paced conditions, participants were asked to complete all walk, jog, and run activities at a pace they deemed appropriate. During the set-paced conditions, participants were told to complete the walk, jog, and run within specified velocities (time frames). Walking was conducted at an average pace of 1-2 $m \cdot s^{-1}$ which equated to a time of 20-40 seconds. Jogging was conducted at an average pace of 3-4 $m \cdot s^{-1}$ which equated to a time of 10-13 seconds. Running was conducted at an average pace of 5-6 $m \cdot s^{-1}$ which equated to a time of 7-8 seconds. During set-paced conditions, participants were told if they were meeting speed requirements at the half-way point. Trials that did not meet these criteria were repeated. An RPE score was taken after each condition (set- and self-paced).

Testing sessions were conducted at the following timeframes for concussed participants. Return to run (RTR), RTP, 2-weeks post RTP, and 4-weeks post RTP. Control participants followed similar testing points, however also completed an additional session at 6-months to assess stability of outcome variables over a prolonged period.

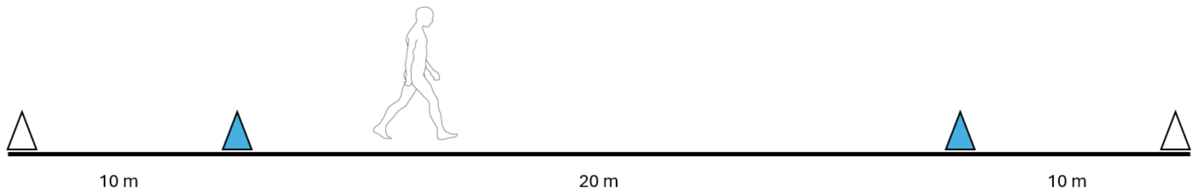


Figure 3.2 Testing protocol utilised for studies within this thesis. White cones represent the beginning and end of the 40-m distance. Blue cones represent the 10-m acceleration and deceleration distances. Participants were instructed to start at the white cone, accelerate to the blue cone, maintain speed across the middle 20-m, decelerate at the second blue cone, stopping at the end white cone.

## 3.4 Gait assessment

### 3.4.1 Gait tasks

Research has been conducted on the utilisation of gait-based assessments to complement current clinical assessments in the management of SRC. Such assessments have incorporated varying gait tasks including single-task, dual-task, and tandem gait variations (114, 151, 156, 244). Previous studies have identified differences in gait patterns following concussion (6, 114, 151, 244). Specific analysis of single-task walking velocities between concussed and control groups have shown significant between group differences ($p = 0.003$), with concussed participants displaying slower walking velocities (266). This is consistent with the analysis of spatiotemporal gait characteristics amongst concussed and control participants. Shorter stride lengths and reductions in gait velocity have been seen amongst concussed individuals during walking trials and are characteristic of a slower and more conservative gait pattern (267).

Currently, research has focused on walking characteristics, with no focus on jogging or running activities. Due to the target population, and the aims of this project, where we wanted to determine whether an easy-to-implement protocol was reliable and valid, a single-task paradigm gait pattern was adopted. Additionally, characteristics of walking, jogging, and running gait were analysed, allowing a more comprehensive analysis of motor strategies during these tasks, and therefore better replicating team sport training and competition demands.

### 3.4.2 Distance

Distances used to assess gait abnormalities vary across studies, ranging from 8- to 20-m (267-269). This project assessed participants over a 40-m total distance, with the first and last 10-m segments incorporated for acceleration and deceleration movements and therefore excluded from analysis. A total of 20-m was used for data analysis, as this allowed a more steady-state pace to be identified, reflecting 'normal' gait, by excluding acceleration and deceleration phases. This distance was also selected to better resemble average distances of standardised team sport warm-ups ranging from 20-30m (270, 271). Further, this allowed for an adequate number of trials (10-14 during the running condition) to be captured to derive variability for the spatiotemporal gait parameters.

## 3.5 Apex STATSports unit – thoracic spine

### 3.5.1 Description

Wearable devices that house both GNSS chips and accelerometers are commonly used in team sports to quantify movement patterns and external load, allowing practitioners to monitor activities completed across the entire playing group (272). Such devices typically consist of a tri-axial accelerometer, measuring accelerations in the antero-posterior, medio-lateral, and vertical axes – typically at 100 Hz, and a global navigation satellite system (GNSS) chip, typically sampling at 10Hz. In addition, these devices may also house gyroscopes and magnetometers to allow for the orientation of the unit in space to be quantified. Based on acceleration and positional information, these devices allow for the quantification of external load, such as speed and distance covered, with the ability to offer insight into fluctuations in activity over time.

The STATSports Apex Pro unit used in this project incorporated a tri-axial accelerometer (100 Hz; ± 16 g), GNSS chip (10 Hz), magnetometer (10 Hz; ± 16 g), and gyroscope (100 Hz; 2000 deg·s$^{-1}$). Aggregated metrics typically reported from these units included; total load (square root of the sum of the triaxial accelerations), total distance and distances in predefined zones, maximum velocity, as well as acceleration and deceleration variables (273, 274). The tri-axial accelerometer also allowed more detailed breakdown of the contribution of all three axes to the total load. This provided greater insight into the movement patterns of individuals. Previous studies investigating the sensitivity of such measures found reductions in vertical acceleration contributions amongst athletes with neuromuscular fatigue (171). Similar findings were reported amongst football players, with changes in movement patterns, as detected by changes in axes contributions, were identified amongst fatigued players (175). These findings support the use of accelerometers to identify subtle changes in movement strategies amongst fatigued athletes, and therefore potentially be used to assess movement strategies in those returning from concussion. In addition, these devices may possess the sensitivity to obtain spatiotemporal parameters of gait such as step count, frequency and variability (162).

Table 3.1 STATSports Apex raw data

| Metric | Description |
|---|---|
| Time | Start and finish time of drill in GMT |
| Latitude | Location of device relative to the equator |
| Longitude | Location of device relative to the Greenwich meridian |
| Speed (m.s$^{-1}$) | Instantaneous velocity (10 Hz) |
| HDOP | Horizontal dilution of precision, which reflects the geometry of the satellites relative to the device |
| Number of satellites | Number of satellites during session used to triangulate position of the device |
| Instantaneous acceleration impulse | Absolute change in speed from the previous data point. |
| Acceleration X | Acceleration in the mediolateral axis |
| Acceleration Y | Acceleration in the vertical axis |
| Acceleration Z | Acceleration in the anterior-posterior axis |
| Gyroscope X | Angular velocity about the mediolateral axis |
| Gyroscope Y | Angular velocity about the vertical axis |
| Gyroscope Z | Angular velocity about the anterior-posterior axis |

### 3.5.2 Validity and reliability

Validity and reliability of accelerometer and GNSS devices are closely related to sampling rates, with higher sampling frequency associated with more accurate and reliable data collection (275). The STATSports Apex unit has a sampling rate of 100 Hz and 10 Hz for the accelerometer and GNSS chip, respectively. Previous studies have noted the potential limitations associated with GNSS units, highlighting a limited ability to capture changes in velocity when compared to other microtechnology devices. Such limitations seem to be exacerbated over short distances during multi-directional activities, such as change of direction and agility tasks (162, 170). However, a review on the validity of GNSS units concluded that devices sampling at 10 Hz were shown to accurately determine total distance, peak velocity, and average speed metrics during straight-line protocols (170). It was noted that protocols involving change of direction, specifically over short distances seemed to reduce the validity of these devices.

Specific analysis of the STATSports Apex units has shown good validity during jogging and sprinting activities. Criterion validity during self-paced jogging activities (20-m and 40-m distance) and sprinting activities (20-m distance) was shown to be acceptable (273). No significant differences between the device and criterion measure were evident. The device displayed an absolute error of $0.62 \pm 0.45$ $km \cdot h^{-1}$, with a bias of $2.36 \pm 1.67\%$ at speeds of $26.5 \pm 2.3$ $km \cdot h^{-1}$ (273). This device has also shown excellent reliability for jogging activities over distances ranging from 20- to 400-m (ICC = 0.96) (273). This is consistent with inter-device reliability for peak running speed over distances ranging from 5- to 30-m (ICC = 0.99)) (274). These findings support the use of the device for straight-line activities over repeated testing points.

Acceleration metrics are typically measured via changes in instantaneous velocity, and are typically categorized into peak, average, or instantaneous acceleration (170). Currently, GNSS devices have shown good ability to quantify average change in pace when using raw data. However, results seem to be compromised when extracting data from manufacturer-based software into discretised acceleration bands (170). Measures of instantaneous velocity were shown to be accurate for straight-line running, however demonstrated poorer ability to quantify during speeds of 1-3 $m \cdot s^{-1}$ (170).

A study assessing the validity of peak resultant accelerations derived from a wearable device found variations in measures during different gait tasks (walk, jog, run) (276). Specifically, as the magnitude of acceleration increased, the device was less capable of accurately measuring peak acceleration, which may be due to saturation of the accelerometer. It was concluded that peak resultant acceleration can be accurately measured during walking and jogging tasks, with questionable validity for running at 5 and 5.9 $m \cdot s^{-1}$, although mean bias was still only 0.18 ± 0.19 G at this speed (276). With regards to data filtering, higher cut-off frequencies (20 Hz) tended to overestimate peak acceleration, while lower frequencies (6 Hz) underestimated. A cut-off of 10 Hz demonstrated greatest accuracy when compared to the criterion (motion analysis) (276, 277).

## 3.6 Vicon Blue Trident inertial measurement units – lumbar spine and ankles

### 3.6.1 Description

Inertial Measurement Units (IMUs) are microtechnology devices that are capable of calculating gait events and spatiotemporal parameters such as stride and step length, stride and step frequency, flight time, and contact time (161, 162). They present the opportunity for clinicians to obtain high-resolution gait characteristics to augment traditional clinical assessments during the management of SRC.

Previously, IMUs have been used to assist with estimation of spatiotemporal characteristics associated with gait amongst clinical populations (278, 279). Sampling frequency of IMUs vary, however a minimum of 25-50Hz sampling rate is recommended for general spatial and temporal gait analysis (280). Higher sampling rates (1600 Hz ± 200 g) will offer higher-resolution analysis of gait characteristics. The IMUs utilised in the current project sample at 1125 Hz ± 16 g on their low g setting and were therefore considered to offer appropriate resolution to measure the stride variables of interest.

Studies have explored the use of IMUs in detecting gait abnormalities following SRC. A single-participant study identified gait impairments (increased step time, step time variability, and step time asymmetry) two-months post-concussion, despite the participant being cleared for return to play (RTP) 20-days post-concussion (268). Similar findings were found amongst adolescents who were between three and 30 ± 22 days post-concussion. Significant differences in medio-lateral sway and gait velocity during a walking task were identified between the two testing points (281). Implementation of dual-task walking trials have also identified differences in gait metrics, specifically slower velocity and increases in postural sway (282).

Table 3.2 Raw data from the Blue Trident IMUs

| Metric | Description |
|---|---|
| Acceleration X | Acceleration ($m{\cdot}s^{-2}$) in the vertical axis |
| Acceleration Y | Acceleration ($m{\cdot}s^{-2}$) in the mediolateral axis |
| Acceleration Z | Acceleration ($m{\cdot}s^{-2}$) in the antero-posterior axis |
| Gyroscope X | Angular velocity ($rad{\cdot}s^{-1}$) about the vertical axis |
| Gyroscope Y | Angular velocity ($rad{\cdot}s^{-1}$) about the mediolateral axis |
| Gyroscope Z | Angular velocity ($rad{\cdot}s^{-1}$) about the antero-posterior axis |
| Magnetometer X | Magnetic orientation in vertical axis |
| Magnetometer Y | Magnetic orientation in mediolateral axis |
| Magnetometer Z | Magnetic orientation in antero-posterior axis |
| High g acceleration X | Collected at sampling frequency of 1600 Hz and range of ±200g |
| High g acceleration Y | Collected at sampling frequency of 1600 Hz and range of ±200g |
| High g acceleration Z | Collected at sampling frequency of 1600 Hz and range of ±200g |
| Low g acceleration X | Collected at sampling frequency of 1125 Hz and range of ±16g |
| Low g acceleration Y | Collected at sampling frequency of 1125 Hz and range of ±16g |
| Low g acceleration Z | Collected at sampling frequency of 1125 Hz and range of ±16g |

The direction of X, Y, and Z axes are determined based on the device being situated in an upright position (e.g. lumbar-mounted). Axis direction will change when the device is rotated (e.g. ankle-mounted)

### 3.6.2 Validity and reliability

The validity of IMU measurements is dependent on their data capture and processing capabilities. Small errors in measurement and the presence of environmental noise in the raw data may lead to drift during integration, thereby reducing accuracy of the measures (280). Sensor orientation is calculated by using the angular velocity measured via the gyroscope, combined with the acceleration data which is used to attempt to correct for any drift (283). The magnetometer also aims to assist in correcting drift by providing magnetic orientation. However, the use of IMUs in indoor environments, where magnetic fields can be distorted, may reduce their ability to use the magnetometer data to correct such errors (283). Drift errors and noise from mechanical motions are typically corrected via filtering and smoothing techniques applied to the raw data.

A recent meta-analysis on the validity and reliability of IMUs in detecting gait metrics concluded that lumbar spine, tibia, and foot placements were valid sensor locations allowing valid measures to be determined as they did not differ from the criterion (high speed camera, force plate, 3D motion capture, instrumented treadmill and/or gait systems) (162). Specifically, measures of step time, stride time, flight time, step frequency, and step length displayed good validity for foot mounted devices when compared to photocell or high-speed camera criterion (162). Tibia placement displayed good validity for measures of stride time. Lumbar placement demonstrated good reliability for measures of stride time, step frequency, and step length during running tasks. However poor validity for flight time, swing time, and step length during sprinting tasks (162). Measures of stride time derived from thoracic mounted IMUs showed significant bias when compared to force plate (162). It was concluded that data derived from positions on the upper body may be limited, and a need to manually classify events may be necessary (162).

Previous studies have shown mixed reliability for IMU devices and placement (162). Inter-session test-retest reliability ranged from poor to excellent for metrics of contact time (foot and lumbar) (ICC = 0.24-0.96), flight time (foot and lumbar) (ICC = 0.94-0.98), step length (foot), and step frequency (foot and lumbar) (ICC = 0.78-0.94) (162). Metrics derived from foot or tibia placements were, on average, most reliable. A study looking at the inter-device reliability of software derived step metrics from the 'IMU Step' desktop application, displayed good reliability for metrics including step count, low intensity steps, medium intensity steps, high intensity steps, impact load, and bone stimulus (284). Inter-unit reliability, on average, demonstrated excellent reliability for most metrics during linear and non-linear running tasks (ICC = 0.90-0.98). Inter-session test-retest reliability of these devices has also shown good to excellent reliability (ICC = 0.73-0.97) during straight line acceleration and deceleration tasks (285). Between day test-retest reliability of a lumbar-mounted Blue Trident IMU during a running-based experimental protocol demonstrated good to excellent reliability (ICC > 0.75) for raw unfiltered acceleration data from the three axes (X,Y,Z) and resultant acceleration signals (286).

## 3.7 Data collection and processing

### 3.7.1 Data collection

Once participants were fitted with devices, they completed the standardised warm-up prior to the testing protocol. Data from the STATSports Apex unit were collected over the duration of the set- and self-paced conditions. Data from the Blue Trident units were collected as separate sessions for set- and self-paced conditions. Trials were counterbalanced and randomised for set-and self-paced conditions. Upon completion, the Blue Trident IMU session was ended via the iPad application, and the STATSports Apex unit was switched off via the device button.

### 3.7.2 Data download

Following data collection, the thoracic-mounted devices were connected to a computer running proprietary software (Sonra 2.0, Newry, Co. Down, N. Ireland) to download the data. Individual drills were then created for each activity (walk, jog, run), trial number (i.e., walk 1), and condition (set- or self-paced). The start and end of each drill were determined from the start and stop of movement shown by the vertical acceleration trace. Once all drills were created, the extended raw data (including drills and whole session) were downloaded to a .csv file and saved in a password protected folder relative to the participant. The raw data were exported as a continuous time series in 100 Hz files, with the 6 channel IMU data and instantaneous velocity provided. A separate .csv file containing the software derived metrics was also downloaded and saved in a password protected folder.

Raw data obtained from the Vicon Blue Trident IMUs was downloaded using proprietary software (Capture.U, Oxford United Kingdom). Data for each condition (set- or self-paced) were downloaded as a separate csv. file and saved in a password protected folder.

### 3.7.3 Data processing

A visual representation of the data processing steps is depicted in Figure 3.3. Pre-processing and analysis of the Apex and Blue Trident data were performed using the R Statistical Programming Language (R Foundation for Statistical Computing, Vienna, Austria) within the R Studio Integrated Development Environment (Posit, Boston, Massachusetts).

Velocity and acceleration data for both devices were filtered using a $4^{th}$ order low-pass Butterworth filter, with the following cut-off frequencies:

- Velocity 1 Hz filter
- Linear accelerations 20 Hz filter

This was done using the *signal* package by first stipulating the functions of the filter using the *butter* function before passing the time series through the filter using the *filtfilt* function. 10 Hz velocity data has been traditionally processed this way (287). A 20 Hz filter was shown to be the best at maximizing IMU worn resultant accelerations compared to a criterion motion capture system (288). Data from the lumbar mounted IMU was processed at 20 Hz, as this has been previously used for similar tasks (289).

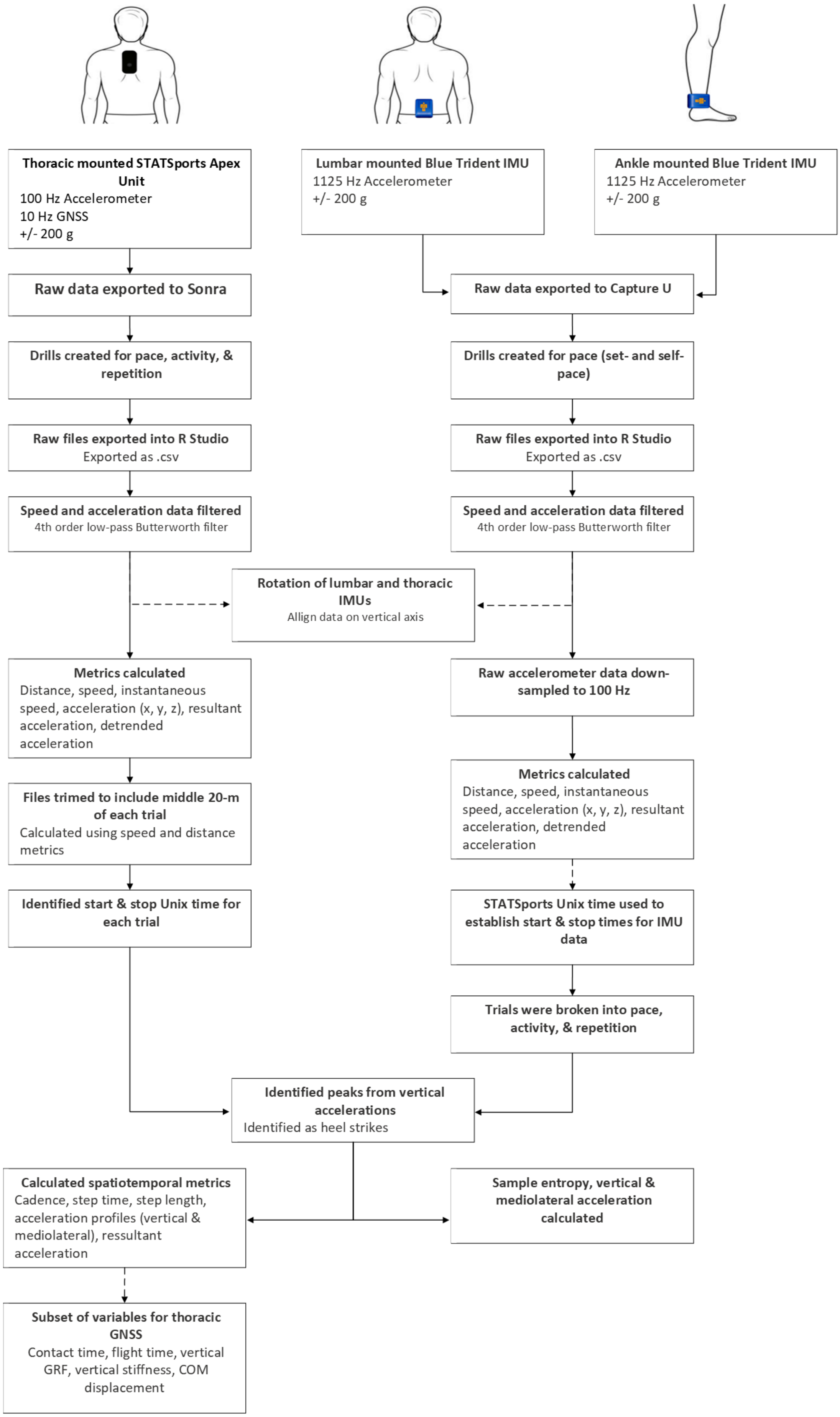


Figure 3.3 Data processing flow chart depicting steps taken during data collection, data download, data processing, and feature extraction.

### 3.7.3.1 Rotation

As the GNSS device only provided linear accelerations in the raw export (*x*, *y*, *z*), to align gravity onto the vertical axis for the thoracic and lumbar mounted units, data were rotated using a 3D rotation matrix. First, a normalised gravity vector was created by taking the average acceleration across the time series for each axis to estimate the direction of gravity. Second, by using the *atan2* function, the angles needed to rotate the accelerometer data were established for each axis. Third, separate rotation matrices for each axis were created using the angles created in step two with the standard formulas for 3D rotation before being combined into a single rotation matrix by multiplying each rotation matrix in order using *dplyr* syntax (*%*%*). Raw data were then passed through this matrix and aligned on the vertical axis. Prior to analysis, acceleration data from all devices were then detrended to remove gravity.

### 3.7.3.2 Down-sample

Accelerometer data obtained from the Blue Trident IMUs worn at the lumbar and ankle region were down sampled from 1125 Hz to 100 Hz to improve processing speed, and to allow direct comparison with accelerometer data obtained from STATSports Apex units. This was done by generating a new time series from the minimum and maximum UNIX times of the original time series and creating appropriate increments to reflect 100 Hz data. Then the original acceleration data were down sampled from 1125 to 100 Hz using the *resample* function in the *signal* package.

### 3.7.3.3 Step detection

Steps were calculated by identifying peaks in vertical acceleration data using the *stat_peaks* function in R, where each peak represented a heel strike. This was performed for each individual file and plotted using function *ggplot* in the *tidyverse* package within R. Arguments within this function allow for an ignore threshold. All data derived from each device had a threshold of 60% of the peak acceleration within the time series to detect a peak. A span argument was set based on 100Hz data and anticipated foot falls, based on the movement velocity. Thoracic- and lumbar-mounted IMUs had a span of 51 for walk activities and 35 for jog and run activities. Ankle-mounted devices had a span of 101 for walk activities and 71 for jog and run activities. Once each step had been detected, the *ggplot* data were extracted to obtain the time that each step occurred along with the magnitude of the vertical acceleration. This data frame containing all the steps and when they occurred in time was joined back onto the main data frame using the *semi_join* function in *dplyr*. The first 4 and last 4 steps of each trial were removed to account for acceleration and deceleration; all other steps were retained to allow spatiotemporal gait variables to be calculated.

3.7.3.4 Start, stop times

Individual drills for each pace condition and repetition were created for thoracic-mounted devices through the Sonra software using the vertical acceleration profiles. Capture U software did not allow for the accurate creation of individual drills; therefore lumbar and ankle data were exported only based on pace condition.

To calculate individual start and stop times for each drill, Blue Trident IMU data was joined to the STATSports Apex data-frame. Raw data was joined using filename, date, and Unix time. The start and stop Unix times previously identified for the trimmed thoracic data (STATSports Apex) was then used to determine start and stop times for the Blue Trident IMU data. Data that did not match with the previously determined start and stop times was excluded.

## 3.8 Feature extraction

A range of outcome measures were analysed to assess potential movement strategy changes following SRC. Outcome variables assessed between group differences over time. Additionally, we assessed for any differences between set- and self-paced conditions. It was hypothesised that, due to slower velocities associated with concussed individuals (156), these participants may show greater variability in gait during set-paced running trials due to the greater speed requirements.

### 3.8.1 Spatiotemporal variables

Averages and standard deviations for a range of spatiotemporal variables were calculated using accelerations derived at each identified vertical peak (heel strike). Variables included number of steps (identified via peak detection), cadence (steps / time (s)), step length (distance at current step − distance at previous step), step time (time at current step − time at previous step), and the acceleration profiles that occurred at each step (antero-posterior, medio-lateral, vertical). Variables were calculated using acceleration data derived at each heel contact (representative of steps). Additional variables; contact time (CT), flight time (FT), vertical ground reaction force (GRF), vertical stiffness (Kvert), and centre of mass (COM) displacement were calculated by extracting x, y, and z variables from thoracic mounted devices using validated algorithm (290).

Metrics derived from the ankle-mounted devices better represented stride and GRF measures due to their proximity to the ground, therefore providing more information on foot-contact movements. Metrics derived from the lumbar spine better represented COM movements, thus providing more information regarding the body's sway during gait movements.

### 3.8.2 Time series analysis

#### 3.8.2.1 Aggregate measures

Analysis of the entire time series was performed to generate several commonly used metrics for each trial, excluding acceleration and deceleration phases of the movements. Average and standard deviation for the resultant acceleration ($\sqrt{(x^2) + (y^2) + (z^2)}$) for each individual trial were calculated. Maximum and average speed ($m \cdot s^{-1}$) were also derived from the thoracic-mounted GNSS device.

#### 3.8.2.2 Non-linear analysis

Entropy is a type of non-linear analysis used to quantify the regularity of the time series by assessing the predictability of a given pattern to reappear based on the provided data points (73, 291-293). Human gait, although chaotic, is a relatively predictable cyclic pattern (294). Measures of entropy can therefore be used to differentiate between healthy and impaired gait by assessing reproducibility of the motor pattern (131, 295-297). A lower entropy score is associated with higher predictability and regularity of movements, whereas higher sample entropies reflect lower predictability or greater irregularity.

This approach may provide additional insight into motor strategy variability that is otherwise missed during more traditional linear measures (298). Approximate entropy was first introduced to assess regularity of heart rate and biological data (292, 299). However, it was established that approximate entropy has a statistical bias when used to assess predictability of discrete data types, notably gait (299). Therefore, sample entropy was introduced. Both sample entropy and the Lyapunov exponent (LyE) have become the common form of non-linear analysis of human data, particularly gait (298, 299). This thesis used sample entropy to assess regularity of gait amongst concussed and control groups. Previous studies utlising sample entropy within concussion management have identified alterations in postural sway during acute phases of concussion, indicating the possible advantages of using non-linear forms of signal analysis (236, 300). Sample entropy was shown to be stable for discrete data lengths of 500 to 5000 data points (299). Within this study, each run consisted of ~700-800 data points, with walks being comprised of ~2500-4000 data points. Previous use of sample entropy examined inter-stride variables derived from discrete moments within a gait cycle (e.g. heel strike) to establish the regularity of variables such as stride time and length (301). Implementation of sample entropy to assess continuous data (e.g. joint positions over a whole time series) have also been successful (301).

Sample entropy requires three input parameters; length of the time series ($N$), tolerance radius ($r$), and embedding dimension ($m$) (292, 293). Length of time series simply refers to the length of the data set (e.g. number of data points within the sequence). Tolerance radius is used to create a tolerance in which subsequent data points can be counted as similar. This is calculated by multiplying $r$ by the SD of the data series to create the tolerance. Embedding dimension creates a parameter within the vectors as to how many data points are included for comparison (292, 299, 301).

### 3.9 Post-Concussion Symptom Scale (PCSS)

### 3.9.1 Assessment description

Symptom inventories are commonly integrated within concussion management to evaluate post-concussion symptoms, as well as gauge return to play (RTP) readiness (302). The Post-Concussion Symptom Scale (PCSS) (264) is one of many inventories used to assess this outcome (303). The PCSS consists of 22-symptoms, evaluated on a 7-point scale (0 to 6), 0 representing no symptoms, and 6 representing severe symptoms (See Appendix 2). Users are asked to provide a subjective evaluation of their symptoms post-concussion, with a total score index provided at the end. This tool is commonly used throughout the management of SRC, providing medical staff insight into symptom severity irrespective of performance on clinical motor assessments.

### 3.9.2 Implementation within project

This project has implemented the PCSS to assess symptom severity amongst the control and concussed participants. Control participants were required to complete the PCSS prior to the first session and concussed participants completed the PCSS prior to all testing sessions, thus providing an indication of current symptoms. Significant differences have been identified during the initial 7 days following a SRC (304), thus providing a rationale for its implementation in this project. However, it is noted that differences in recovery between age groups may occur, specifically, it was found that high school athletes (under 18-years) demonstrated significant increases in post-concussive symptoms 24-hours, 3-days, and 5-days post injury compared to college athletes who only demonstrated significant differences 24-hours and 3-days post-concussion (305). Due to this, it is anticipated that participants under 18-years of age may have a prolonged recovery period prior to being eligible to complete testing.

Concussed participants were tested at return to run (approximately 3-5 days post), RTP (approximately 10 days post), 2-weeks post RTP, and 4-weeks post RTP. Prior to their participation, potentially eligible participants had their symptom severity assessed in accordance with standardised protocols from sporting organisations and were, therefore, cleared by a medical doctor. Therefore, the implementation of the PCSS within this study served as a precautionary measure, ensuring symptoms did not exceed acceptable levels, nor become more severe between testing periods.

### 3.9.3 Outcome measures

The PCSS is the most common outcome measure used to assess symptom severity. This is calculated by summing severity scores of each symptom, with a total score ranging from 0 – 132 (264). This project only focused on total symptom score as an outcome measure. All concussed participants eligible for this study were cleared by a medical doctor to resume activity, therefore indicating a return to medically acceptable scores for symptom questionnaires and clinical assessments. Therefore, this project will only use the PCSS total score to provide insight into potential symptomology at time of testing.

### 3.9.4 Validity and reliability

The PCSS is a well-validated symptom questionnaire (303, 304, 306, 307). Previous research has identified significant differences in total score approximately 3 days post-concussion when compared to baseline scores (303). This trend is consistent with other research, identifying differences up to 7-14 days post-concussion (304). Although this score has shown high discriminant ability, arguments surrounding the utility of this outcome measure exist. In particular, total score may not provide full insight into symptom patterns, therefore limiting its ability to differentiate between two athletes with the same score (306). Studies have investigated the effectiveness of symptom cluster scores (303). In particular, dividing into four symptom clusters; cognitive, physical, affective, and sleep (303). Results showed significant differences from baseline to post-concussion for global severity index (summary measure), physical symptom cluster, positive symptom score (symptom breadth), and total symptom score (303).

The PCSS has shown high internal reliability amongst healthy and concussed adults, with Cronbach's Alpha of 0.87 and 0.94, respectively (303). Reliability between healthy male and female adolescents and young adults also showed high reliability, with Cronbach's Alpha of 0.88 and 0.91, respectively (307). The standard error of the measurement (SEM) for total scores amongst male (±3.4 points) and female (±4.4 points) populations using 80% confidence intervals (307) suggest the PCSS is a reliable measure of symptom severity. However, differences between sex groups have been previously identified (306, 307), with females commonly reporting higher symptom severity than males. Therefore, it is recommended that male and female participant scores be assessed in their respective groups.

## 3.10 Rating of Perceived Exertion (6-20)

### 3.10.1 Assessment description

The Borg Rate of Perceived Exertion (RPE) (308) is used to assess subjective interpretation of effort, exertion, and breathlessness during physical activity (309). Initially designed to rate exertion during dynamic work, this scale also provides both practitioners and participants with subjective quantification of training load. The scale ranges from 6-20, 6 representing no exertion, and 20 representing maximal exertion (308, 309). The numerical order of the scale is based on scores being linearly aligned with heart rate (HR), and is shown to have high correlations between the two (309). For example, an RPE of 6 correlates with a HR of approximately 60 beats per minute.

### 3.10.2 Implementation within project

This project implemented the Borg 6-20 RPE scale throughout all testing periods. Previous findings showed that acutely concussed individuals reported greater RPE scores during exercise when still symptomatic (310). This was evident despite exercising at lower intensities. During the testing protocol, participants were asked to rate their RPE after both set- and self-paced conditions, providing insight into perceived exertion during both trials. Participants were familiarised with the RPE scale prior to testing via verbal description. Previous studies have shown that prior familiarisation of the scale does not affect outcome scores (311), however may ensure participants can differentiate between perceived exertion and other physiological responses to avoid over- or under-estimating the score (312). The RPE scale was divided into coloured bands with brief descriptions to help differentiate intensity bands (See Appendix 3). This has been previously validated amongst younger and older adult populations (313).

### 3.10.3 Outcome measures

Perceived exertion defined as the numerical rating on the 6- to 20-point scale is the sole outcome measure for this assessment. As per previous studies, it is expected that RPE scores collected after 5-minutes of aerobic exercise for set-and self-paced conditions will be positively associated with average HR. This score will be recorded on an excel spread sheet for each testing session and referred to during data analysis.

### 3.10.4 Validity and reliability

The Borg RPE scale has demonstrated high criterion validity between RPE score, HR and/or HR reserve (311, 314, 315). Studies have shown significant correlations between RPE score and HR ($r = 0.70$, $p = <0.001$) particularly during walking-based dynamic exercise (311). These findings are also consistent with measures of HR reserve, showing high correlations between the two measures ($r = 0.69$) during an intermittent cycle test (314). Other studies have demonstrated correlations between RPE score and other physiological measures, including cardiac frequency (CF) during walking-based incremental ($r = 0.58$) (316) and continuous ($r = 0.85$) (317) tests at different intensities. Therefore, these results support the use of the Borg RPE scale within this project, highlighting its ability to correlate with physiological exertion during walking-based exercise.

Test-retest reliability of this assessment is demonstrated through consistency across repeated trials. Studies investigating this have found moderate test-retest reliability (ICC = 0.79) for the repeated use of the Borg 6-20 RPE scale amongst older adults (315). Additionally, high test-retest reliability (ICC = 0.78) for RPE at ventilatory threshold was also found amongst children (318). Therefore, these findings support the repeated use of the Borg RPE scale amongst adult and adolescent populations.

# Chapter 4

# Reliability of Ankle, Lumbar and Thoracic Mounted IMUs

Reliability of wearable devices for measuring gait parameters on grassed surfaces amongst recreationally active adults

This study aims to be published in the Journal of Sports Sciences.

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Please refer to Appendix 4 for ethics approval and Appendix 5 for a copy of the participant information letters and consent forms.

## 4.1 Abstract

*Background:* The assessment of gait is of interest for clinical and sporting populations to determine movement capabilities for performance profiling or following return to play from injury. It is not clear whether gait can be reliably assessed on grassed surfaces at various velocities using commercially available inertial measurement units (IMUs) at various anatomical sites. *Aims:* The aims of this study were to assess the test-retest reliability of ankle, lumbar and thoracic mounted IMUs for deriving spatiotemporal and nonlinear gait metrics during walking, jogging and running tasks performed on grassed surfaces. *Methods:* Using a counterbalanced, randomised, repeated measures design, 29 participants (20 males, age = 25.5 ± 4.8 years; nine females, age = 23.4 ± 3.7 years) completed four testing sessions over a 5-week period. Participants were fitted with a thoracic-mounted wearable device, which housed a global navigation satellite system (GNSS) chip and triaxial IMU (Apex Pro Series, STATSports) and three triaxial IMU devices: one on the medial aspect of each ankle, and at the approximate location of the 4$^{th}$ to 5$^{th}$ lumbar vertebrae. Participants completed three × 40-m walk, jog, and run tasks under set- and self-paced conditions on a natural grassed football field. Relative and absolute test-retest reliability were quantified via intraclass correlation coefficients (ICC), standard error of measurement (SEM), and minimal detectable change expressed in absolute and relative units (MDC). *Results:* Most variables across all movement velocities displayed good to excellent relative reliability with ICCs greater than 0.75. Cadence, step time average, resultant acceleration, and vertical ground reaction force (GRF) were most reliable. Step time variation and mediolateral acceleration displayed poor relative reliability (ICC = 0.3 – 0.86). All anatomical sites, movement velocities and set- and self-paced conditions displayed comparable reliability, with measures being slightly more reliable at the thoracic and lumbar spine. For absolute reliability, the average normalised minimal detectable change (MDC) was 22 ± 18% across all variables and conditions, highlighting most variables were sensitive to detecting changes in movement strategies. *Conclusions:* In general, reliable gait metrics can be obtained from IMUs during velocities ranging from walking to high-speed running on grassed surfaces. Thoracic mounted devices could be used during self-paced, constant velocity tasks during regular training to obtain these measures. Practitioners can use the MDCs from this study to track performance changes in athletes.

**Keywords:** Test-retest reliability, inertial measurement unit, gait, sports-related concussion

## 4.2 Introduction

Individual gait patterns, like signatures, are unique and are controlled by the central nervous system (CNS) (319). Locomotion requires coordination between multiple domains stemming from the CNS. The coordinated interaction between muscle activation and kinematic joint position provides the ability to maintain cyclical relationships between body segments, whilst simultaneously maintaining postural stability (320, 321). Given the repeatability of gait, it can be used as a tool to assess changes in CNS function in controlled environments, such as a laboratory (322).

Assessment of gait is common in the management of many clinical populations (301, 320, 323-326). Previously, such assessments have relied on laboratory equipment such as 3D motion analysis or pressure sensitive walkways (325-328), which limits the uptake of these methods beyond clinical practice. The capability and availability of portable technologies, such as inertial measurement units (IMUs), have made these assessments more feasible and accessible for practitioners (251, 319, 324). Inertial measure units consist of a tri-axial accelerometer, gyroscope, and magnetometer measuring accelerations, angular velocity, and magnetic field strength (to orient data with the Earth's axes), respectively (162, 327, 329). Previous studies have demonstrated the ability of these devices to objectively quantify gait amongst healthy and clinical populations within controlled environments (320, 323, 324, 327).

There are several instances where monitoring gait via IMUs is important within athletic populations such as monitoring neuromuscular fatigue (179), following lower limb musculoskeletal injuries (e.g. ACL reconstruction) (330, 331), and following sports-related concussion (SRC) (74, 110, 332, 333). Within athletes with SRC, there appears to be an increased risk of musculoskeletal injury following return to play (RTP), suggesting sub-clinical abnormalities exist beyond RTP (3-6, 76). Athletes competing on natural grass surfaces have also been shown to sustain a subsequent musculoskeletal injury sooner when compared to synthetic surfaces (e.g. artificial turf) (334). Indeed, concussed athletes tend to show more conservative yet variable gait, with more variable step time, slower cadence and movement speed, and longer time in double support compared to healthy controls (8, 115, 151, 154, 169, 246-250, 335). Current clinical management of SRC relies heavily on subjective clinical assessments (4, 70, 73) alongside a graduated RTP protocol, which is governed by a re-integration into physical activities limited by exacerbation of symptoms. Multifaceted clinical assessments, like the Sport Concussion Assessment Tool (SCAT) or the Sport Concussion Office Assessment Tool 6 (SCOAT6) (336), assess motor deficits via the modified Balance Error Scoring System (mBESS) and timed tandem gait (134). Within these tests the assessment of complex tandem gait, dual-task gait, and graded aerobic assessment are optional for practitioners to complete (336). These motor assessments, although demonstrating moderate to good reliability, these assessments have shown poor sensitivity, particularly beyond one-week post-concussion (337). Given that laboratory assessments are challenging in the field, and subjective assessments may not offer suitable resolution, potentially IMUs potentially represent a more cost-effective, everyday solution for assessing movement changes within routine management following SRC (161) and in the everyday monitoring of athletes.

Foot, lumbar, and thoracic mounted IMUs have been shown to possess good reliability for variables such as; contact time, flight time, step frequency, vertical stiffness (Kvert), and vertical ground reaction force (GRF) across a range of velocities (2.2 $m.s^{-1}$ - 8.0 $m.s^{-1}$) and distances (40m – 400m) when performed on firm surfaces (162, 179). Furthermore, measures of vertical ground reaction force displayed excellent reliability when derived from foot or ankle mounted devices during treadmill, outdoor, and indoor gait tasks (162).

Thoracic mounted wearable technology, incorporating both global navigation satellite systems (GNSS) and IMU devices, have been extensively used in sport as a means of monitoring the external loads of athletes (170). Accumulated accelerometer load per minute and vertical accelerations have been used to identify changes in movement strategies under player fatigue (171, 173, 175). These alterations may imply impaired capacity for the CNS to produce efficient gait under fatigued conditions (171, 173, 175). Given that concussed athletes demonstrate changes in gait, thoracic wearable technology may provide an avenue to objectively and routinely examine gait in athletes who are recovering from SRC.

Given the widespread use of thoracic mounted devices, determining the reliability for deriving metrics that are useful in the assessment of sub-clinical neuromuscular changes post-concussion and various markers of neuromuscular performance is warranted. The aims of this study were to assess the test-retest reliability of ankle, lumbar and thoracic mounted IMUs during walking, jogging and running tasks performed on grassed surfaces. Results from this study will assist in determining the efficacy of IMUs for assessing gait characteristics in the field and their potential use in identifying practically meaningful changes following SRC.

## 4.3 Methods

### 4.3.1 Participants

A total of 31 healthy, recreationally active participants were recruited (21 males, 10 females). Of the 31 participants who completed the initial assessment, 2 participants withdrew from the study following the initial session due to unforeseen commitments. A total of 29 (20 males, age 25.5 ± 4.8; 9 females, age 23.4 ± 3.7) participants were included in the study. Of the included participants, 12 male participants and two female participants recorded a history of previous concussion. Participants had no diagnosed concussion within 12-months of testing. This was reliant on athlete recall and therefore did pose limitations in reliability (338). Using the *ICC. Sample.Size* package in R and the *calculateSampleSize* function, with the minimum acceptable intraclass correlation coefficient (ICC) of 0.5 and an expected ICC of 0.7, a sample size of 29 participants were required to achieve 80% power (339). Across the 29 participants, a total of 112 sessions were performed, with 27 participants completing all four testing sessions (initial, one week post, three weeks post, and five weeks post initial). To encourage natural gait, all participants wore their normal training footwear which was consistent across each session for each participant. Participants were free from any neurological disorders, had no lower limb injuries within three months of testing, and no diagnosed concussions within 12-months of testing. Participants were provided with an online information letter and provided e-consent if they volunteered for the study. All procedures were approved by the university's human research ethics committee (2021-214H).

### 4.3.2 Study design and procedures

This study used a counterbalanced, randomised, repeated measures design. Reliability of the devices was assessed using data collected on four separate occasions. Testing sessions were designed to replicate time frames for individuals returning to play (RTP) following a SRC. Times included initial testing (replicating return to run), one-week post initial (replicating RTP), two-weeks post RTP, and four-weeks post RTP. The protocol was performed on natural grassed surfaces to replicate the conditions under which players would perform following RTP. Similarly, to better replicate the movement characteristics required for athletic populations, all participants performed walking, jogging, and running tasks.

Participants were fitted with a thoracic-mounted wearable device (Apex Pro Series, STATSports) which housed a triaxial accelerometer (100 Hz; ± 16 g), GNSS chip (10 Hz), magnetometer (10 Hz; ± 16 G), and gyroscope (100 Hz; ± 2000 $deg \cdot s^{-1}$) positioned in an upright orientation in a tight-fitting manufacturer provided vest at the approximate level of the third and fourth thoracic vertebrae. Accelerations along the antero-posterior (Z), medio-lateral (X), and vertical (Y) axes were recorded. Participants used the same device for each testing session. Participants were also fitted with three Blue Trident IMU devices with a triaxial accelerometer (± 16 g) sampling at 1125 Hz, magnetometer (100 Hz), and gyroscope (1125 Hz; 2000 $deg \cdot s^{-1}$) (version 10.0.1, VICON, Oxford, UK). Specifically, the three IMUs were positioned on the medial aspect of each ankle using manufacturer provided straps, and on the skin overlying the spine at the approximate height of the fifth lumbar vertebrae using double-sided tape reinforced with strapping tape over the top of the IMU. The X, Y, and Z axes for the lumbar-mounted device represented vertical, medio-lateral, and antero-posterior directions, respectively. Due to placement of the ankle mounted units, orientation of the X, Y, and Z axes shifted to antero-posterior, vertical, and medio-lateral, respectively. All devices were fitted and calibrated in line with the manufacturer guidelines prior to each testing session. The thoracic-mounted device was switched on approximately 10-minutes prior to testing, to obtain a suitable satellite signal. The GNSS devices used have been shown to offer suitable reliability and validity for measuring instantaneous velocity during straight-line movements (170, 265).

Prior to completing the protocol, participants completed a standardised warm-up consisting of 10-m jog, 10 × squats, lunges, and hamstring sweeps. Participants were then instructed to complete three × walks, jogs, and runs under set- and self-paced conditions over a 40-m distance. Whilst the order of activities was the same across sessions, trials were counterbalanced and randomised to complete either the set- or self-paced condition first. Participants were instructed to complete each activity from a static starting position.

During the self-paced conditions, participants were asked to complete all walk, jog, and run activities at their own pace. During the set-paced conditions, participants were required to complete the walk, jog, and run tasks within certain time frames. Walking was conducted at a pace of 1 to 2 $m \cdot s^{-1}$ which equated to a time of 20- to 40-seconds. Jogging was conducted at a pace of 3 to 4 $m \cdot s^{-1}$ which equated to a time of 10- to 13-seconds. Running was conducted at a pace of 5 to 6 $m \cdot s^{-1}$ which equated to a time of 7- to 8-seconds. During set-paced conditions, participants were told if they were meeting speed requirements at the half-way point. Trials that did not meet times were repeated. A Rating of Perceived Exertion (RPE) score was taken after each condition (set- and self-paced) using the Borg RPE 6-20 scale (340).

### 4.3.3 Data processing

Following each testing session, the thoracic-mounted devices were connected to a computer running proprietary software (Sonra 2.0, Newry, Co. Down, N. Ireland). Individual drills were then created for each activity (walk, jog, run), repetition (i.e., walk 1), and condition (set- or self-paced). The start and end of each drill were determined by marking on the vertical accelerometer timeseries when movement began and ceased. Once all drills were created, the extended raw data (including drills and whole sessions) were exported to a comma-separated values (.csv) file. The raw data were exported as a continuous time series in 100 Hz files, with the 6 channel IMU data and instantaneous velocity provided. A separate .csv file containing the software derived metrics was also exported for each session. Raw data obtained from the Blue Trident IMUs were downloaded using proprietary software (Capture.U, Oxford, United Kingdom). Data for each condition (set- or self-paced) was exported as separate .csv files and saved in a separate password protected folder. Once the IMU data from each location were downloaded, all subsequent preprocessing and analysis of data were performed using the R Statistical Programming Language (R Foundation for Statistical Computing, Vienna, Austria) within the R Studio Integrated Development Environment (Posit, Boston, Massachusetts).

Data from the GNSS and IMUs were filtered using a 4th order low-pass Butterworth filter; velocity and linear accelerations were filtered using 1 Hz and 20 Hz filters, respectively (286, 287). This was done using the *signal* package by first stipulating the functions of the filter using the *butter* function before passing the time series through the filter using the *filtfilt* function. The ankle and lumbar data were then down sampled from their original 1125 Hz to 100 Hz to allow for comparisons with the thoracic mounted IMU data. This was done by generating a new time series from the minimum and maximum UNIX times of the original time series and creating appropriate increments to reflect 100 Hz data. Then the original acceleration data were down sampled from 1125 to 100 Hz using the *resample* function in the *signal* package.

As the GNSS device only provided linear accelerations in the raw export (*x*, *y*, *z*), to align gravity onto the vertical axis for the thoracic and lumbar mounted units, data were rotated using a 3D rotation matrix. First, a normalised gravity vector was created by taking the average acceleration across the time series for each axis to estimate the direction of gravity. Second, by using the *atan2* function, the angles needed to rotate the accelerometer data were established for each axis. Third, separate rotation matrices for each axis were created using the angles created in step two with the standard formulas for 3D rotation before being combined into a single rotation matrix by multiplying each rotation matrix in order using *dplyr* syntax (*%*%*). Raw data were then passed through this matrix and aligned on the vertical axis. The average acceleration on each axis for the thoracic prior to rotation was x = -0.016 g, y = 0.852 g, and z = 0.509 g, following rotation this was x = 0.00 g, y = 1.01 g, and z = 0.01. Prior to analysis, acceleration data from all devices were then detrended to remove gravity.

### 4.3.4 Step detection

A peak detection algorithm was used to identify heel strikes within the vertical acceleration data of each individual file. Identification of peaks was performed using the *stat_peaks* function in R. Data were plotted using the *ggplot* function in the *ggplot2* package within R. Arguments within this function allow for an ignore threshold on the y-axis and a span on the x-axis. A threshold of 60% of the peak acceleration within the time series was set to detect a peak and a span was set based on 100Hz data and anticipated foot falls, based on the movement velocity.

All plots were exported and then visually inspected to ensure peaks were appropriately detected. Any issues identified were individually corrected through changing span parameters.

### 4.3.5 Feature extraction

A range of spatiotemporal variables were calculated using accelerations derived at each identified vertical peak (heel strike). Variables included, number of steps (identified via peak detection), cadence (steps / time (s)), average step time (time / number of steps), and acceleration profiles (antero-posterior, medio-lateral, vertical). Average and standard deviation for the resultant acceleration $\sqrt{((x^2) + (y^2) + (z^2))}$ over the time series for each trial were also calculated. Sample entropy, a non-linear analysis, was performed on accelerations, allowing the regularity of a time series to be quantified by assessing how well the current and past characteristics of a time series can predict future characteristics of the same time series (341). Additional variables; contact time (CT), flight time (FT), vertical GRF, vertical stiffness (Kvert), and centre of mass (COM) displacement were calculated for jog and run activities by extracting x, y, and z variables from thoracic mounted devices using validated algorithm for jogging and running tasks (290) (Table 4.1).

Table 4.1 List of linear and non-linear outcome measures

| **Measure** | **Location** |
|---|---|
| *Linear* | |
| Speed (m/s) | Thoracic |
| Contact time (s) | Thoracic |
| Flight time (s) | Thoracic |
| Vertical ground reaction force (kN) | Thoracic |
| Vertical stiffness (Kvert) | Thoracic |
| COM displacement (cm) | Thoracic |
| Cadence (steps.$s^{-1}$) | Thoracic, lumbar, ankle |
| Step time average (s) | Thoracic, lumbar, ankle |
| Step time variation (sd) | Thoracic, lumbar, ankle |
| Acceleration, vertical | Thoracic, lumbar, ankle |
| Acceleration, mediolateral | Thoracic, lumbar, ankle |
| Resultant acceleration | Thoracic, lumbar, ankle |
| *Non-linear* | |
| Sample entropy, vertical acceleration | Thoracic, lumbar, ankle |
| Sample entropy, mediolateral acceleration | Thoracic, lumbar, ankle |

### 4.3.6 Statistical analyses

All statistical analyses were conducted in RStudio (version 2023.06.1+524, Posit, Boston, MA, USA) using R programming language (version 4.2.3 R Foundation for Statistical Computing, Vienna, Austria). To determine relative and absolute test-retest reliability of outcome variables from each device location (ankle, lumbar, thoracic), each activity (walk, jog, run) and each pace condition (self- and set-pace), several statistical tests were performed. Firstly, relative reliability over time of each outcome variable was assessed using a two-way mixed-effects ICC model to calculate intraclass correlation coefficients (ICC) (version 2,1) across time points, with single, absolute agreement (342). This was performed using the *icc* function in the *irr* package. Thresholds used to interpret the correlations were poor (< 0.50), moderate (0.50 – 0.75), good (> 0.75 – 0.90), and excellent (> 0.90) reliability (342). Reliability was deemed acceptable at an ICC ≥ 0.70 based on the COnsensus-based Standards for the selection of health Measurement INstruments (COSMIN) guidelines for assessing the quality of studies on reliability and measurement error of outcome measurement instruments (194).

To determine absolute reliability and thresholds required to interpret changes over time, standard error of the measurement (SEM), minimal detectable change (MDC); representing the smallest change that can be detected beyond random error, and normalised MDC (rMDC) expressed as a percentage were calculated (343).

$$\text{SEM} = \text{standard deviation first test} \times \sqrt{1} - \text{ICC}$$

$$\text{MDC} = 1.96 \times \text{SEM} \times \sqrt{2}$$

$$\text{rMDC (\%)} = (\text{MDC / mean}) \times 100$$

An additional analysis was conducted on all variables with pace conditions combined. Additional variables: mediolateral acceleration, and sample entropy, mediolateral acceleration was added. Absolute reliability, SEM, and MDC were calculated and are presented in Table 4.3. The full R code used for the analysis is shown in Appendix 6.

## 4.4 Results

### 4.4.1 Thoracic mounted units

As shown in Table 2, all variables displayed acceptable relative reliability with moderate to excellent ICCs (*r* = 0.62-0.98) across set-paced and self-paced conditions and all activities. Measures of cadence, average step time, sample entropy, resultant acceleration and vertical GRF were most consistent across set-paced and self-paced conditions and all activities. For absolute reliability, all normalised MDCs were below 50% other than for vertical stiffness (Figure 4.1); most variables required changes of less than 30% relative to the mean to represent a detectable change, with an average of 21 ± 14%.

As shown in Table 3, when pace conditions were combined, most variables displayed acceptable relative reliability with *moderate* to *excellent* ICCs (*r* = 0.64-0.97) for all activities. Measures of mediolateral acceleration displayed *poor* to *moderate* ICCs (*r* = 0.34-0.71). For absolute reliability, all normalised MDCs were below 50% other than for mediolateral acceleration.

### 4.4.2 Lumbar mounted units

As shown in Table 4.2, cadence, step-time average, sample entropy, and resultant acceleration displayed acceptable relative reliability with moderate to good ICCs (*r* = 0.71-0.9). Measures of step time variation displayed poor to good reliability (*r* = 0.36-0.79), with poorer reliability associated with increasing velocity. For absolute reliability, all normalised MDCs were below 50%, other than for step time variation, with the average being 25 ± 24% (Figure 4.1).

As shown in Table 4.3, when pace conditions were combined, most variables displayed acceptable relative reliability with *moderate* to *good* ICCs (*r* = 0.58-0.87) for all activities. Measures of mediolateral acceleration and step time variation displayed *poor* to *moderate* ICCs (*r* = 0.43-0.75). For absolute reliability, all normalised MDCs were below 50% other than for mediolateral acceleration and step time variation.

### 4.4.3 Ankle mounted units

As shown in Table 4.2, cadence, step-time average, resultant acceleration, and sample entropy displayed acceptable relative reliability with moderate to good ICCs ($r$ = 0.54-0.88). All measures other than sample entropy were not affected by set-paced and self-paced conditions or activities. Sample entropy displayed poorer reliability as velocity increased. Step time variation displayed poor to moderate ICCs ($r$ = 0.4-0.72). For absolute reliability, all normalised MDCs were below 50% other than for step time variation (Figure 4.1); most variables required changes less than 30% relative to the mean, average 22 ± 16%.

As shown in Table 4.3, when pace conditions were combined, most variables displayed acceptable relative reliability with *moderate* to *good* ICCs ($r$ = 0.52-0.87) for all activities. Measures of mediolateral acceleration and step time variation displayed *poor* to *moderate* ICCs ($r$ = 0.3-0.64). For absolute reliability, all normalised MDCs were below 50% other than for mediolateral acceleration and step time variation.

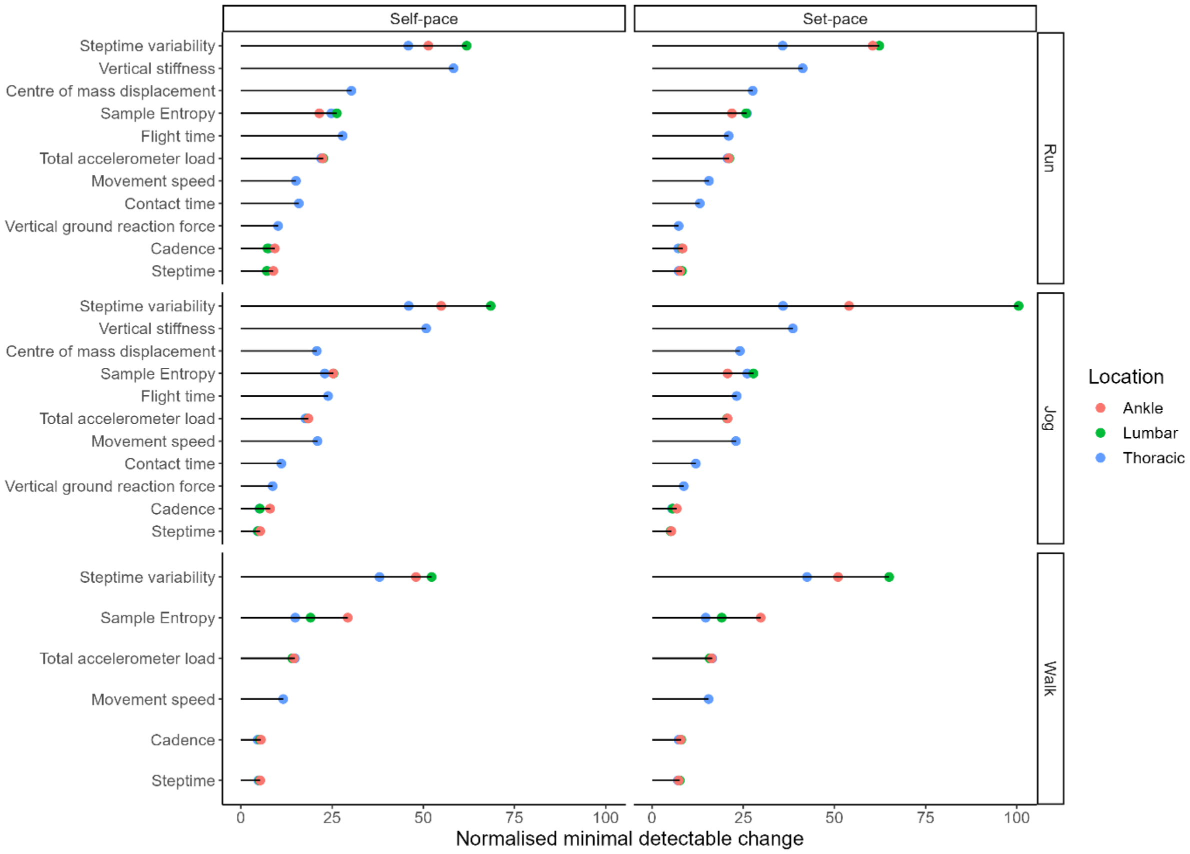


Figure 4.1 Normalised minimal detectable change for ankle- (red), lumbar- (green), and thoracic-mounted (blue) devices during set- and self-paced walk, jog, and run activities. Values represent the change required to determine a real change in performance. This is relative to the mean and expressed as a percentage.

Table 4.2 Summary of ICC, SEM, and MDC for thoracic, lumbar, and ankle mounted devices during each activity and pace condition

| | **Walk** | | | | | | **Jog** | | | | | | **Run** | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | Set-Pace | | | Self-Pace | | | Set-Pace | | | Self-Pace | | | Set-Pace | | | Self-Pace | | |
| | ICC | SEM | MDC | ICC | SEM | MDC | ICC | SEM | MDC | ICC | SEM | MDC | ICC | SEM | MDC | ICC | SEM | MDC |
| Thoracic | | | | | | | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.84 [0.75-0.91] | 0.05 | 0.14 | 0.91 [0.86-0.95] | 0.03 | 0.09 | 0.85 [0.77-0.92] | 0.06 | 0.16 | 0.87 [0.8-0.93] | 0.05 | 0.15 | 0.84 [0.76-0.91] | 0.09 | 0.24 | 0.82 [0.72-0.9] | 0.09 | 0.25 |
| Step time variation (sd) *(s)* | 0.85 [0.77-0.92] | 0.01 | 0.02 | 0.86 [0.78-0.92] | 0 | 0.01 | 0.86 [0.78-0.92] | 0 | 0.01 | 0.78 [0.66-0.87] | 0 | 0.01 | 0.81 [0.71-0.89] | 0 | 0.01 | 0.69 [0.56-0.82] | 0 | 0.01 |
| Step time average *(s)* | 0.85 [0.76-0.91] | 0.01 | 0.04 | 0.91 [0.86-0.95] | 0.01 | 0.03 | 0.87 [0.79-0.92] | 0.01 | 0.02 | 0.89 [0.82-0.94] | 0.01 | 0.02 | 0.85 [0.76-0.91] | 0.01 | 0.02 | 0.84 [0.75-0.91] | 0.01 | 0.02 |
| Sample entropy, vertical | 0.87 [0.79-0.92] | 0.02 | 0.06 | 0.89 [0.83-0.94] | 0.02 | 0.06 | 0.75 [0.63-0.85] | 0.04 | 0.11 | 0.81 [0.72-0.89] | 0.03 | 0.1 | 0.7 [0.57-0.82] | 0.06 | 0.15 | 0.76 [0.64-0.86] | 0.05 | 0.14 |
| Resultant acceleration *($m{\cdot}s^{-2}$)* | 0.83 [0.74-0.9] | 9.9 | 27.44 | 0.83 [0.74-0.9] | 8.82 | 24.45 | 0.8 [0.7-0.89] | 32.1 | 88.98 | 0.86 [0.78-0.92] | 27.64 | 76.63 | 0.86 [0.79-0.92] | 31.85 | 88.28 | 0.88 [0.81-0.93] | 33.66 | 93.3 |
| Speed *($m{\cdot}s^{-1}$)* | 0.69 [0.55-0.81] | 0.08 | 0.21 | 0.76 [0.64-0.86] | 0.06 | 0.16 | 0.62 [0.47-0.76] | 0.32 | 0.89 | 0.76 [0.65-0.86] | 0.28 | 0.78 | 0.7 [0.56-0.82] | 0.29 | 0.81 | 0.79 [0.69-0.88] | 0.27 | 0.75 |
| Contact time *(s)* | | | | | | | 0.73 [0.61-0.84] | 0.01 | 0.03 | 0.73 [0.61-0.84] | 0.01 | 0.03 | 0.7 [0.57-0.82] | 0.01 | 0.03 | 0.59 [0.44-0.74] | 0.01 | 0.04 |
| Flight time *(s)* | | | | | | | 0.63 [0.48-0.77] | 0.01 | 0.03 | 0.77 [0.66-0.87] | 0.01 | 0.03 | 0.69 [0.56-0.81] | 0.01 | 0.02 | 0.59 [0.44-0.74] | 0.01 | 0.03 |
| Vertical GRF *(N)* | | | | | | | 0.97 [0.95-0.98] | 0.06 | 0.17 | 0.97 [0.95-0.98] | 0.06 | 0.18 | 0.98 [0.96-0.99] | 0.05 | 0.15 | 0.97 [0.95-0.98] | 0.07 | 0.18 |
| Vertical stiffness (Kvert) *($kN{\cdot}m^{-1}$)* | | | | | | | 0.73 [0.6-0.84] | 3.79 | 10.5 | 0.69 [0.55-0.81] | 4.83 | 13.4 | 0.67 [0.53-0.8] | 5.33 | 14.77 | 0.65 [0.5-0.78] | 5.97 | 16.54 |
| COM displacement *(m)* | | | | | | | 0.71 [0.58-0.83] | 0.01 | 0.02 | 0.73 [0.61-0.84] | 0.01 | 0.02 | 0.7 [0.56-0.82] | 0.01 | 0.02 | 0.61 [0.46-0.75] | 0.01 | 0.02 |
| Lumbar | | | | | | | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.82 [0.71-0.9] | 0.05 | 0.15 | 0.89 [0.81-0.94] | 0.03 | 0.1 | 0.8 [0.69-0.89] | 0.06 | 0.16 | 0.85 [0.76-0.92] | 0.05 | 0.15 | 0.75 [0.62-0.86] | 0.09 | 0.26 | 0.83 [0.73-0.91] | 0.08 | 0.23 |
| Step time variation (sd) *(s)* | 0.72 [0.58-0.84] | 0.01 | 0.03 | 0.79 [0.67-0.88] | 0.01 | 0.02 | 0.36 [0.19-0.56] | 0.01 | 0.02 | 0.57 [0.4-0.73] | 0.01 | 0.01 | 0.48 [0.31-0.67] | 0.01 | 0.02 | 0.48 [0.31-0.66] | 0.01 | 0.02 |
| Step time average *(s)* | 0.83 [0.73-0.91] | 0.02 | 0.04 | 0.9 [0.83-0.95] | 0.01 | 0.03 | 0.84 [0.74-0.91] | 0.01 | 0.02 | 0.87 [0.79-0.93] | 0.01 | 0.02 | 0.75 [0.63-0.86] | 0.01 | 0.03 | 0.84 [0.74-0.91] | 0.01 | 0.02 |
| Sample entropy, vertical | 0.71 [0.58-0.83] | 0.04 | 0.11 | 0.75 [0.62-0.85] | 0.04 | 0.11 | 0.74 [0.61-0.84] | 0.05 | 0.14 | 0.74 [0.61-0.84] | 0.04 | 0.12 | 0.76 [0.65-0.86] | 0.06 | 0.16 | 0.75 [0.63-0.86] | 0.06 | 0.16 |
| Resultant acceleration *($m.s^{-2}$)* | 0.84 [0.76-0.91] | 9.5 | 26.35 | 0.85 [0.76-0.91] | 8.38 | 23.23 | 0.8 [0.7-0.89] | 32.05 | 88.83 | 0.85 [0.76-0.91] | 29.03 | 80.46 | 0.86 [0.78-0.92] | 32.77 | 90.84 | 0.88 [0.8-0.93] | 34.64 | 96.02 |
| Ankle | | | | | | | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.81 [0.71-0.89] | 0.03 | 0.07 | 0.85 [0.77-0.92] | 0.02 | 0.05 | 0.74 [0.61-0.85] | 0.04 | 0.11 | 0.75 [0.62-0.85] | 0.04 | 0.12 | 0.8 [0.69-0.89] | 0.05 | 0.15 | 0.76 [0.64-0.86] | 0.06 | 0.16 |
| Step time variation (sd) *(s)* | 0.4 [0.23-0.59] | 0.01 | 0.02 | 0.55 [0.39-0.72] | 0.01 | 0.01 | 0.55 [0.38-0.71] | 0.01 | 0.02 | 0.49 [0.33-0.67] | 0.01 | 0.01 | 0.56 [0.39-0.72] | 0.01 | 0.02 | 0.72 [0.59-0.84] | 0.01 | 0.02 |
| Step time average *(s)* | 0.83 [0.74-0.91] | 0.03 | 0.09 | 0.88 [0.81-0.93] | 0.02 | 0.06 | 0.81 [0.71-0.89] | 0.02 | 0.04 | 0.84 [0.74-0.91] | 0.01 | 0.04 | 0.83 [0.73-0.91] | 0.02 | 0.05 | 0.77 [0.66-0.87] | 0.02 | 0.06 |
| Sample entropy, vertical | 0.75 [0.66-0.82] | 0.02 | 0.06 | 0.79 [0.72-0.85] | 0.02 | 0.06 | 0.73 [0.65-0.81] | 0.04 | 0.1 | 0.59 [0.48-0.7] | 0.05 | 0.13 | 0.54 [0.43-0.66] | 0.04 | 0.12 | 0.58 [0.47-0.69] | 0.04 | 0.12 |
| Resultant acceleration *($m.s^{-2}$)* | 0.83 [0.77-0.88] | 9.8 | 27.15 | 0.84 [0.78-0.89] | 8.61 | 23.87 | 0.8 [0.73-0.86] | 32.34 | 89.65 | 0.84 [0.79-0.89] | 29.05 | 80.52 | 0.86 [0.81-0.91] | 32.37 | 89.71 | 0.88 [0.83-0.92] | 34.29 | 95.05 |

ICC, intraclass correlation coefficient along with the [95% confidence interval]; SEM, standard error of the measure; MDC, minimal detectable change; sd, standard deviation; vGRF, vertical ground reaction force; Kvert, vertical stiffness; COM, centre of mass; $m{\cdot}s^{-1}$, meters per second; $m{\cdot}s^{-2}$, meters per second squared; N, Newtons, $kN{\cdot}m^{-1}$, kilonewton per meter

Table 4.3 Extended Summary of ICC, SEM, MDC, and normalised MDC for thoracic, lumbar, and ankle mounted devices during each activity with set- and self-paced conditions combined

| **Variable** | **ICC [CIs]** | **SEM** | **MDC** | **rMDC (%)** | **ICC [CIs]** | **SEM** | **MDC** | **rMDC (%)** | **ICC [CIs]** | **SEM** | **MDC** | **rMDC (%)** |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | **Walk** | | | | **Jog** | | | | **Run** | | | |
| Thoracic | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.87 [0.82-0.91] | 0.04 | 0.11 | 6 | 0.86 [0.81-0.91] | 0.06 | 0.15 | 5.2 | 0.83 [0.77-0.88] | 0.09 | 0.24 | 7.4 |
| Step time variation (sd) *(s)* | 0.85 [0.8-0.9] | 0.01 | 0.01 | 38.4 | 0.81 [0.74-0.87] | 0 | 0.01 | 38.2 | 0.75 [0.67-0.83] | 0 | 0.01 | 38.1 |
| Step time average *(s)* | 0.88 [0.83-0.92] | 0.01 | 0.03 | 6 | 0.88 [0.83-0.92] | 0.01 | 0.02 | 4.8 | 0.84 [0.78-0.89] | 0.01 | 0.02 | 7.2 |
| Acceleration vertical ($m \cdot s^{-2}$) | 0.84 [0.78-0.89] | 0.05 | 0.14 | 21.5 | 0.83 [0.77-0.88] | 0.22 | 0.6 | 18.6 | 0.86 [0.81-0.91] | 0.21 | 0.6 | 16.9 |
| Acceleration mediolateral ($m \cdot s^{-2}$) | 0.71 [0.61-0.79] | 0.01 | 0.03 | 62 | 0.36 [0.25-0.49] | 0.1 | 0.28 | 97.1 | 0.34 [0.23-0.47] | 0.12 | 0.33 | 94.9 |
| Resultant acceleration ($m \cdot s^{-2}$) | 0.83 [0.77-0.88] | 9.37 | 25.98 | 15.2 | 0.83 [0.77-0.88] | 30.06 | 83.33 | 18.6 | 0.87 [0.82-0.91] | 33 | 91.47 | 20.7 |
| Sample entropy, vertical | 0.88 [0.83-0.92] | 0.02 | 0.06 | 14.4 | 0.78 [0.7-0.85] | 0.04 | 0.1 | 23.3 | 0.73 [0.64-0.81] | 0.05 | 0.15 | 24.7 |
| Sample entropy, mediolateral | 0.95 [0.92-0.96] | 0.03 | 0.08 | 13.5 | 0.74 [0.66-0.82] | 0.05 | 0.13 | 17 | 0.69 [0.59-0.78] | 0.06 | 0.16 | 18.3 |
| Speed ($m \cdot s^{-1}$) | 0.7 [0.61-0.79] | 0.07 | 0.19 | 13.5 | 0.7 [0.61-0.79] | 0.3 | 0.84 | 20.6 | 0.75 [0.67-0.83] | 0.28 | 0.79 | 15.1 |
| Contact time *(s)* | | | | | | | | | | | | |
| Flight time *(s)* | | | | | 0.71 [0.62-0.79] | 0.01 | 0.03 | 22.9 | 0.64 [0.53-0.74] | 0.01 | 0.02 | 24.6 |
| Vertical GRF *(N)* | | | | | 0.97 [0.96-0.98] | 0.06 | 0.18 | 9 | 0.97 [0.96-0.98] | 0.06 | 0.17 | 8.7 |
| Vertical stiffness (Kvert) ($kN \cdot m^{-1}$) | | | | | 0.7 [0.61-0.79] | 4.43 | 12.29 | 45.2 | 0.66 [0.56-0.75] | 5.64 | 15.63 | 49.5 |
| COM displacement *(m)* | | | | | 0.72 [0.63-0.8] | 0.01 | 0.02 | 22.2 | 0.65 [0.55-0.75] | 0.01 | 0.02 | 29 |
| Lumbar | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.84 [0.78-0.89] | 0.05 | 0.13 | 6.7 | 0.83 [0.76-0.88] | 0.06 | 0.16 | 5.3 | 0.79 [0.71-0.86] | 0.09 | 0.25 | 7.7 |
| Step time variation (sd) *(s)* | 0.75 [0.66-0.83] | 0.01 | 0.03 | 54 | 0.43 [0.3-0.56] | 0.01 | 0.02 | 76.4 | 0.48 [0.36-0.61] | 0.01 | 0.02 | 55.3 |
| Step time average *(s)* | 0.86 [0.8-0.91] | 0.01 | 0.04 | 6.5 | 0.85 [0.79-0.9] | 0.01 | 0.02 | 4.8 | 0.79 [0.72-0.86] | 0.01 | 0.03 | 7.6 |
| Acceleration vertical ($m \cdot s^{-2}$) | 0.8 [0.72-0.86] | 0.06 | 0.17 | 25.6 | 0.71 [0.61-0.8] | 0.31 | 0.86 | 27.2 | 0.67 [0.57-0.77] | 0.33 | 0.91 | 26.9 |
| Acceleration mediolateral ($m \cdot s^{-2}$) | 0.58 [0.46-0.7] | 0.05 | 0.13 | 85.9 | 0.44 [0.32-0.58] | 0.26 | 0.71 | 89.6 | 0.52 [0.4-0.64] | 0.25 | 0.7 | 86.8 |
| Resultant acceleration ($m \cdot s^{-2}$) | 0.85 [0.79-0.89] | 8.95 | 24.8 | 14.5 | 0.82 [0.76-0.88] | 30.69 | 85.08 | 18.9 | 0.87 [0.81-0.91] | 34.02 | 94.29 | 21.2 |
| Sample entropy, vertical | 0.72 [0.64-0.8] | 0.04 | 0.11 | 18.5 | 0.72 [0.64-0.8] | 0.05 | 0.13 | 25.7 | 0.76 [0.68-0.83] | 0.06 | 0.16 | 25 |
| Sample entropy, mediolateral | 0.75 [0.66-0.82] | 0.04 | 0.12 | 17.7 | 0.6 [0.49-0.7] | 0.08 | 0.21 | 26.8 | 0.58 [0.47-0.69] | 0.06 | 0.17 | 21 |

Table 4.3 continued

| | | | | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Ankle | | | | | | | | | | | | |
| Cadence *(steps/seconds)* | 0.83 [0.79-0.87] | 0.02 | 0.06 | 6.7 | 0.75 [0.69-0.81] | 0.04 | 0.11 | 7.3 | 0.78 [0.73-0.83] | 0.06 | 0.16 | 8.8 |
| Step time variation (sd) *(s)* | 0.44 [0.36-0.53] | 0.01 | 0.02 | 44.7 | 0.47 [0.38-0.56] | 0.01 | 0.02 | 49.1 | 0.6 [0.52-0.68] | 0.01 | 0.03 | 50.5 |
| Step time average *(s)* | 0.85 [0.81-0.89] | 0.03 | 0.07 | 6.3 | 0.83 [0.79-0.87] | 0.01 | 0.04 | 5.3 | 0.8 [0.75-0.85] | 0.02 | 0.06 | 8.3 |
| Acceleration vertical *($m{\cdot}s^{-2}$)* | 0.85 [0.8-0.88] | 0.25 | 0.69 | 18.6 | 0.79 [0.74-0.84] | 0.46 | 1.27 | 20.8 | 0.79 [0.73-0.84] | 0.52 | 1.43 | 18.6 |
| Acceleration mediolateral *($m{\cdot}s^{-2}$)* | 0.3 [0.21-0.39] | 0.29 | 0.8 | 81.4 | 0.64 [0.56-0.71] | 0.45 | 1.24 | 72.8 | 0.45 [0.37-0.54] | 0.58 | 1.62 | 82.3 |
| Resultant acceleration *($m{\cdot}s^{-2}$)* | 0.83 [0.79-0.87] | 9.25 | 25.65 | 15 | 0.82 [0.78-0.86] | 30.78 | 85.32 | 19 | 0.87 [0.83-0.9] | 33.63 | 93.21 | 21 |
| Sample entropy, vertical | 0.77 [0.71-0.82] | 0.02 | 0.06 | 28.4 | 0.66 [0.59-0.73] | 0.04 | 0.11 | 22.4 | 0.56 [0.48-0.64] | 0.04 | 0.12 | 21.3 |
| Sample entropy, mediolateral | 0.64 [0.57-0.71] | 0.06 | 0.16 | 27.1 | 0.6 [0.52-0.67] | 0.06 | 0.17 | 24.8 | 0.52 [0.44-0.61] | 0.07 | 0.19 | 26.6 |

ICC, intraclass correlation coefficient along with the [95% confidence interval]; SEM, standard error of the measure; MDC, minimal detectable change; rMDC, normalised MDC; sd, standard deviation; vGRF, vertical ground reaction force; Kvert, vertical stiffness; COM, centre of mass; $m{\cdot}s^{-1}$, meters per second; $m{\cdot}s^{-2}$, meters per second squared; N, Newtons, $kN{\cdot}m^{-1}$, kilonewton per meter

## 4.5 Discussion

The primary aim of this study was to investigate the test-retest reliability of ankle, lumbar, and thoracic mounted devices during overground walking, jogging, and running tasks amongst healthy young adults on grassed surfaces. In general, most outcome measures displayed moderate to excellent reliability for gait activities performed on natural grassed surfaces. Test-retest reliability was optimal for metrics derived from the lumbar and thoracic spine. If shown to be sensitive to clinically relevant indications, these findings support the use of IMUs to assess gait characteristics in an applied setting, such as following SRC or as a general athlete monitoring tool. Cadence, step time and resultant acceleration were the most reliable variables across the different locations and activities (e.g., walk) whilst step-time variation was the least reliable. In general, pace condition (set-paced and self-paced) and movement velocity did not impact reliability. Normalised MDC was equal to or below 30% of the mean for all measures other than step time variation, indicating a relatively small change is required to determine a real change in performance for most variables. Collectively, these results demonstrate that gait characteristics are consistent over a 5-week period, with measures of cadence, average step time, resultant acceleration, and vertical GRF being the most reliable. These devices could be utilised to track changes in important gait variables and measures of movement rhythmicity over time at various velocities.

This study showed that spatiotemporal measures of gait can be reliably assessed across all activities and pace conditions. At the thoracic spine, these variables displayed excellent relative reliability, with similar results seen at the lumbar spine and ankle. The exception to this was step time variability, which displayed the poorest reliability of all variables. Measures of step time variation have consistently demonstrated poor reliability, with many studies highlighting the implications associated with measures of variability during gait tasks, in particular during overground walking (344, 345) or on uneven surfaces (346, 347) due to the inherent variations in ground surface. Limitations surrounding sample size may also impact the reliability of step time variation. Due to the subtle variations in gait speed seen in healthy gait (170, 348), obtaining a large enough sample of continuous gait cycles is required to account for this (349, 350), which limits the utility of this variable in many practical settings. With regards to this study, it is plausible to suggest that the distances used within the protocol did not allow a sufficient number of continuous gait cycles to be recorded, thus providing an inaccurate representation of step time variability (349, 350). Across all spatiotemporal variables, cadence, and average step time displayed the best reliability across all activities, locations and pace conditions, with normalised MDC being less than 10%. These findings are consistent with walking tasks performed on treadmills or indoor surfaces using lumbar (279, 351-353) and ankle mounted devices (279, 354).

Another novel aspect of this study was the inclusion of non-linear metrics, in this case, sample entropy, vertical and sample entropy, mediolateral. Both measures displayed *moderate* to *excellent* reliability. Measures of sample entropy, mediolateral were, in general, less reliable. Notably, reliability was reduced as velocity increased. Sample entropy requires sufficient data points to accurately quantify the regularity of the time-series (293). It is plausible to assume that the length of time-series data obtained from this study, specifically as velocity increases (e.g. less steps in data set) may impact the reliability. Determining whether reliability is improved with more data to measure sample entropy should be explored. Previously, non-linear metrics have shown promise in quantifying the regularity of human gait for healthy and clinical (e.g. ageing) populations (298, 301, 355). Measures of entropy have previously been used to differentiate between healthy and impaired gait by assessing the reproducibility of the movement pattern (131, 295-297, 301, 355). This information assists clinicians determining falls and injury risk amongst these populations. With regards to concussion, studies have used measures of entropy to assess differences between healthy and concussed individuals during static and dynamic balance tasks (356-359). Sample entropy has also been used to predict relative risk of sustaining a SRC amongst rugby athletes performing a dynamic balance assessment (236). Due to the complexity of gait, the addition of non-linear analysis, such as sample entropy, may provide practitioners with an added layer of information better informing them of athletes' movement patterns. Collectively, these results show that most spatiotemporal and nonlinear gait metrics can be quantified with a high degree of precision on natural grassed surfaces.

Within a sporting context, measures of cadence have been used to determine differences in gait strategies amongst concussed athletes when compared to controls or pre-concussion results (110, 332, 333). However, there is some speculation as to whether measures of cadence are sensitive enough, with changes typically only seen during dual-task conditions as opposed to single-task (74, 110, 332, 333). Therefore, practitioners need to be aware that reliance on cadence during single-task gait may not be a substantial or clinically relevant outcome measure. From a performance perspective, measuring spatiotemporal variables (like cadence) alongside running velocity could be useful when monitoring changes in sprint performance in athletes. Similarly, resultant acceleration, which can be used as a gross measure of external load, showed excellent reliability (165-167), which is in accordance with others (167). However, as it is a gross measure, it may not provide sufficient resolution to detect changes in movement strategies or characteristics which may occur when the individual axes from the IMU are assessed to detect fatigue (175, 360) or post-concussion abnormalities (6, 361).

A subset of variables were derived from thoracic mounted devices for jog and run activities that have been found to be valid and reliable during jogging and running tasks on an artificial turfed surface (290). These variables included contact time (time between heel strike and toe off), flight time (time between toe off and heel contact of consecutive steps), vertical GRF, Kvert, and COM displacement. Whilst all variables displayed moderate to excellent ICCs and good absolute reliability, vertical GRF and COM displacement displayed the best reliability across jog and run activities from these variables. The reliability of these measures is comparable to the outcomes reported in the initial study where the tasks were performed on an artificial surface (290). Measures of vertical GRF may aid in assessing asymmetries between limbs as well as lower limb stiffness (362). In concussed cohorts, athletes have demonstrated alterations in lower limb stiffness (6, 363) and changes in COM displacement within lab-based tasks (266, 267, 364, 365). Measures of contact time and flight time are commonly used to assess running economy (366, 367) and may assist in detecting gait abnormalities following concussion or be used to monitor the performance, neuromuscular fatigue, or RTP from lower limb musculoskeletal injury of players without the need for additional testings sessions (179). These results highlight the ability of these devices to capture important aspects of running performance reliably, allowing practitioners to monitor these variables daily.

When assessing device placement, the results of our study show that thoracic mounted devices were, in general, similar or more reliable across all outcome measures and activities when compared to lumbar and ankle placements. Ankle and lumbar mounted devices have proven to be reliable and highly accurate tools to measure spatiotemporal metrics during gait tasks performed on firm, stable surfaces due to their proximity to the ground (368-371) and centre of mass, respectively (167, 279, 351-353). Reliability of certain spatiotemporal metrics (vGRF, COM displacement, Kvert) are mixed, with some studies reporting good reliability (290), and others displaying mixed reliability (172, 372, 373). The current findings suggest that these anatomical sites can be used to capture gait metrics on grassed surfaces. This study also shows that thoracic mounted devices worn in a tight-fitting garment can also be used. Indeed, thoracic mounted devices may circumvent some of the feasibility issues associated with the other locations. Devices attached at the ankle need to be small enough to avoid discomfort and not interfere with sport-specific tasks. Practitioners also need access to specialised straps to firmly secure devices. Similarly, lumbar mounted devices need specialised equipment or tape to securely affix the IMUs to the body. These issues currently limit the utility of these sites for collection of data during regular training or competition. Given most team sports already utilise thoracic mounted devices, and considering the findings of our study, this may be the most practical and reliable location for practitioners to monitor athletes' gait patterns. Thoracic mounted IMUs have previously been shown to offer reliable gait outcomes while walking and running on firm surfaces (167, 279, 351-353), with the current study building on this to demonstrate reliable gait parameters on grassed surfaces. Whilst they also appear sensitive to detecting fatigue following sprint protocols (179), evaluation of their efficacy in detecting changes following injury or SRC is warranted.

Further to assessing reliability at various sites on the body, this study also investigated the test-retest reliability of IMUs at multiple velocities. It was shown that reliability was not impacted by changes in velocity or whether the velocities were set- or self-paced for most linear and nonlinear variables. Measures of sample entropy, mediolateral were most affected by velocity, with reduced reliability associated with increased velocity. When determining a return to sport post-concussion protocol, practitioners should therefore be aware of; 1) the reliability of outcome measures associated with a device location that meets their capabilities; and 2) the noise associated with that device location, activity, and pace condition to determine what constitutes a practically important change. Being able to collect these variables under self-paced conditions is important in practice as it means practitioners can simply extract periods of constant velocity during training where the velocity may differ between sessions and or players. Further, assessing gait across a range of velocities may be important to understand how gait changes under the different task constraints.

The SEM and MDC values for each outcome variable are also important for practitioners to consider. The SEM represents the standard error of the measure, whereas the MDC represents the minimal change required for a change to be real as opposed to error of the instrument (342). Most outcome measures displayed a relatively low SEM and a greater MDC regardless of activity, pace condition, or device location. Normalised MDC may offer practitioners a more practical and digestible measure to determine real changes in performance. All variables other than step time variation and Kvert (thoracic) required changes equal to or less than 30% of the mean. Given the relatively low normalised MDC values across most variables and conditions, these metrics have a good capacity to detect changes in performance over time. When selecting variables to track performance, it is first important to determine whether the measure is valid and provides meaningful data to inform decision making processes and then understand how much change is required to first be a real change (i.e., >MDC) and then a practically meaningful change.

There are some limitations to this study. Firstly, there was an unequal sex distribution of participants included in the study, with more males than females. Differences in gait patterns may exist between males and females (374) and therefore sex-specific reliability differences may be masked during a pooled analysis. This study was limited in recruitment of female participants, however future research should look to investigate if sex-related differences in gait do exist, and if so, how they may impact reliability outcomes. Secondly, we did not derive measures of contact time, flight time, vertical GRF, vertical stiffness, and COM displacement for lumbar- and ankle-mounted devices. The validated algorithm used to derive these outcomes was based on thoracic-mounted units, therefore limiting our ability to apply this to lumbar- and thoracic-mounted locations. Whilst the IMU in the GNSS unit contains a gyroscope and magnetometer, the manufacturer only provides the linear accelerations to be exported, which meant a sensor fusor algorithm using gyroscopic data was not possible. This approach would have provided a more accurate representation of the device orientation when removing gravitational accelerations. However, given the average acceleration following the rotation being 1 on the vertical axis for the thoracic IMU, we are confident the orientation estimation still provided us with meaningful data. We did not control for firmness of the ground in anyway, which will have varied slightly across trials and participants. However, this may better reflect real-world testing environments. Practitioners should be aware that some of the more discrete variables such as contact time may be more easily detected on firm, consistent surfaces.

Finally, as mentioned, devices at the lumbar region were affixed to the skin via tape and therefore may have been more susceptible to unwanted movement, ultimately influencing the reliability.

## 4.6 Conclusion

This study sought to assess the test-retest reliability of IMUs on grassed surfaces at varying velocities and provides a bridge between laboratory and field-based applications. Spatiotemporal measures can be reliably derived from thoracic, lumbar, and ankle mounted IMUs. Step time variation had the poorest reliability across velocities and device locations. Non-linear analyses using sample entropy, showed acceptable reliability during jog activity for ankle mounted devices, and all activities for thoracic and lumbar mounted devices. Clinicians and practitioners should be aware of the relative feasibility of each device location and select a location that is most practical for them. Device location will then provide clinicians and practitioner with information regarding the best outcome measures and activities to perform. Variables were consistent across set- and self-paced conditions and movement velocities allowing practitioners to reliably obtain these metrics during regular training activities without the need for additional assessments.

## 4.7 Linking paragraph

The findings of this study show that ankle, lumbar, and thoracic-mounted units can reliably measure a variety of linear and non-linear gait metrics during walk, jog, and run activities performed on natural grassed surfaces. Within the context of concussion management, the findings presented within this study may provide clinicians and practitioners the opportunity to assess gait characteristics of athletes within the field and without the need of costly, specialised equipment. Clinicans and practitioners can be confident that gait characteristics can be reliably collected during simple gait tasks of increasing velocities. The simplicity of the assessment allows data to be collected during various straight-line gait tasks performed throughout training or match-play scenarios, therefore eliminating time constraints that are typically associated with other assessments. Future research into the sensitivity of wearable microtechnology to monitor changes post-concussion is warranted. Notably, the collection of gait data from team sport athletes may provide an opportunity for comparison over the duration of a season and/or post-injury (e.g. concussion, lower-limb musculoskeletal). Data collected may potentially provide insight into alterations in gait strategies, prompting further investigation and the potential for more specific rehabilitation programs.

# Chapter 5

# Movement Changes Persist Beyond Return-to-Play During Walking, Jogging and Running Following Sport-related Concussion

Movement changes persist beyond return-to-play during walking, jogging and running following sport-related concussion

This study aims to be published in the Journal of Science and Medicine in Sport.

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Please refer to Appendix 4 for ethics approval and Appendix 5 for a copy of the participant information letters and consent forms.

## 5.1 Abstract

*Background:* Inertial measurement units (IMU) offer a practical solution to measure gait in the field across a range of velocities. They may enable practitioners to regularly assess gait within the field following concussion. *Aims:* The aims of this study were to assess changes in gait using lumbar- and thoracic-mounted IMUs during walking, jogging and running tasks performed on grassed surfaces up to four weeks post return-to-play (RTP) post-concussion. *Methods:* Using a case-control design, 16 medically diagnosed sports-related concussions in 15 team sport athletes were included; 14 males (89.2 ± 13.9 kg, 21.7 ± 3.3 years) and one female (62.5kg, 18 years) alongside healthy age and sex matched controls; 15 males (90.4 ± 12.7 kg, 26.2 ± 5 years) and one female (80kg, 20 years). Participants completed four testing sessions at; return to run, return to play (RTP), 2-weeks post RTP, and 4-weeks post RTP. Participants wore a thoracic-mounted wearable device, which housed a global navigation satellite system (GNSS) chip and IMU, and a lumbar-mounted IMU. Participants completed three × 40-m walk, jog, and run tasks on a grassed football field under set- and self-paced conditions. Spatiotemporal gait metrics and sample entropy were quantified from the IMUs. Bayesian mixed-effects models were used to assess differences between groups over time. Bayes factors were used to help interpret the strength of the evidence. *Results:* The average Post Concussion Symptom Scale (PCSS) score across all sessions was 2.9 ± 5.6 for the concussed group and 3.8 ± 9.9 for the control group. There were no differences in set- and self-paced tasks. There were numerous differences in gait between the concussion and no concussion history groups, with strong to very strong evidence to suggest lower vertical and increased mediolateral accelerations and increased sample entropy at both thoracic and lumbar IMU sites. These differences tended to increase with greater velocities. Step length and cadence showed small differences between groups. Most gait metrics improved over time, typically returning to the level of the control within 40 days post-concussion. The longest recovery times were associated with mediolateral (up to 77-days) and vertical (up to 75-days) sample entropy. Recovery patterns differed depending on velocity, with vertical sample entropy displaying little to no improvement for walk and jog activities. Cadence, mediolateral acceleration, and mediolateral sample entropy displayed no improvement when walking *Conclusions:* Lumbar- and thoracic-mounted IMUs show differences in gait post-concussion which varies depending on movement velocity. There were still observable differences at 60 days post-concussion in some gait metrics. Instrumented gait assessments over a range of velocities should be used post-concussion to monitor athletes. We highlight that commercially available IMUs can be

used to in the field to readily perform gait assessments with athletes returning from concussion. Results may provide practitioners with additional information, informing them of individualised return-to-play timeframes or rehabilitation strategies.

## 5.2 Introduction

Sports related concussions (SRC) are common in both contact (e.g., tackle) and non-contact (e.g., struck by ball) sports (375, 376). Contact sports with a greater number of collisions present with highest rates of SRC (15, 17, 183). Rugby league and Australian rules football both display high incidence, with approximately 14.7 and 4.2-9.4 concussions per 1000 playing hours, respectively (22, 377). At youth levels, rugby, hockey, and American football have demonstrated high incidence amongst youth athletes, with rates respectively reported as 4.18, 1.2, and 0.53 per 1000 athletic exposures (17).The diagnosis and symptom monitoring of SRC relies primarily on multifactorial, subjective clinical assessments, such as the Sport Concussion Assessment Tool 6 (SCAT6), assessing cognitive, motor, and self-report symptoms (378). Within these assessments, the modified Balance Error Scoring System (mBESS) and timed tandem gait are used to assess motor performance, with the number of errors and time to completion used as the outcome variables. Average recovery times based on these assessments typically range from one to four weeks (2, 3, 64, 69, 70, 379).

Standard RTP protocols for most athletes include a minimum stand down period, which may vary by sport and playing level. For example, in Australia, professional rugby league uses 11-days, whereas amateur sport has a 21-day period (82). Within this period, players complete a six-stage graded RTP process (183, 380) that allows athletes to progress to the next stage based on no symptom exacerbation (183, 378). Some sports require athletes to meet pre-injury baseline scores for assessments within the SCAT6, mBESS, and timed tandem gait (378). Of concern, however, is the lack of evidence supporting the use of these diagnostic tools (e.g., SCAT6) beyond the acute stages of concussion, particularly when assessing motor changes. There is emerging evidence for an increased risk of subsequent concussion and musculoskeletal injury up to 12-months post-concussion (3-6), suggesting subclinical changes in function persist beyond RTP. Laboratory assessments have highlighted this, with impairments in static balance (209, 229, 230, 232) and walking gait (145, 239, 335) seen up to three-months post-concussion. However, these assessments lack ecological validity and equipment costs mean that they are largely prohibitive for practitioners.

Microtechnology sensors, notably inertial measurement units (IMUs) are portable devices that allow a range of three-dimensional movement-related metrics. IMUs consist of a tri-axial accelerometer, gyroscope, and magnetometer that measure linear acceleration, angular velocity, and magnetic field strength (to orient data with the Earth's axes), respectively (162, 327, 329). These devices may offer a solution for the assessment of common spatiotemporal gait metrics outside of the laboratory setting following SRC. They have displayed good to excellent reliability, specifically for a range of spatiotemporal metrics, such as step frequency (162), cadence (279, 351-353), and resultant acceleration (167). Additionally, lumbar- and thoracic-mounted units have displayed moderate to excellent reliability during walking, jogging, and running tasks on grassed surfaces (381). With regards to concussion, the use of lumbar- and/or foot-mounted IMUs during walking tasks when performed in controlled environments have shown promise in differentiating between concussed and control participants up to two-months post-concussion (115, 145, 151, 169, 247, 335, 379). Variability in acceleration and peak acceleration metrics, increased sway volume, reduced cadence, or increased time to complete task have been shown post-concussion (115, 145, 151, 169, 247, 335, 379). Whilst these instruments highlight that differences in gait persist beyond RTP, they have still only assessed walking gait and use systems that are not typically used by practitioners in the field.

Thoracic mounted wearable devices that house both a global navigation satellite system (GNSS) chip and IMU are routinely worn by team sport athletes to monitor external loads (170, 171, 173, 175) and appear to be able to capture key gait characteristics such as cadence and other more discrete measures like centre of mass (COM) displacement (290). They have also been used to monitor neuromuscular fatigue amongst players through variations in IMU-derived metrics, including vertical and medio-lateral accelerations (171, 173, 175, 179). Therefore, thoracic-mounted devices may provide similar advantages when assessing movement changes post-concussion with the added innovation of being able to monitor these changes outside of laboratory settings. The aim of this study was to assess changes in spatiotemporal and nonlinear gait metrics using lumbar- and thoracic-mounted IMUs during walking, jogging, and running performed on natural grassed surfaces up to four weeks post RTP. Results from this study will provide much needed insight into the future management of SRCs. It was hypothesised that concussed athletes would present with altered gait strategies characterised by slower, more conservative gait, greater mediolateral acceleration contributions, and less repeatable gait patterns. It was expected that alterations would be exacerbated across activities of increasing velocity, with improved outcomes seen as days since concussion increased.

## 5.3 Methods

### 5.3.1 Participants

A total of 15 concussed participants with 16 medically diagnosed concussion were included in this study; 14 males (89.2 ± 13.9 kg, 21.7 ± 3.3 years) and 1 female (62.5kg, 18 years). One player sustained two concussions 4.5 months apart. Participants were eligible if they sustained a SRC diagnosed by a medical practitioner in-line with the most recent consensus statement at the time when the concussion was diagnosed (187, 378). Eligible participants completed their first assessment once they were cleared to return to run. One player sustained two concussions 4.5 months apart. A total of 16 age and sex matched controls were included in the study, 15 males (90.4 ± 12.7 kg, 26.15 ± 5 years) and 1 female (80kg, 20 years). Eligible participants were required to be free of any neurological disorders and have had no lower limb injuries within three months of testing. In addition, eligible control participants must not have had a diagnosed concussions within 12-months of testing. Participants were provided with an online information letter and provided e-consent if they volunteered for the study. All procedures were approved by the university's human research ethics committee (2021-214H) and in accordance with the Declaration of Helsinki (382).

### 5.3.2 Study design and procedures

This study used a counterbalanced, case-control, repeated measures design and used the same protocol described in chapter four. Prior to each testing session, concussed and control participants completed the post-concussion symptom scale (PCSS) (264), a 21-item, 7-point Likert scale assessment of symptom severity. Participants were fitted with a thoracic-mounted wearable device (Apex Pro Series, StatSports) which housed a triaxial accelerometer (100 Hz; ± 16 g), GNSS chip (10 Hz), magnetometer (10 Hz; ± 16 G), and gyroscope (100 Hz; ± 2000 deg·s$^{-1}$) positioned in an upright orientation in a tight-fitting manufacturer provided vest at the approximate level of the third and fourth thoracic vertebrae. Participants were also fitted with three Blue Trident IMU devices with a triaxial accelerometer (± 16 g) sampling at 1125 Hz, magnetometer (100 Hz), and gyroscope (1125 Hz; 2000 $deg{\cdot}s^{-1}$) (version 10.0.1, VICON, Oxford, UK) on the medial aspect of each ankle and at the approximate location of the fifth lumbar vertebra using double-sided tape. Participants completed a standardised warm-up prior to testing. Participants were then instructed to complete three × walks, jogs, and runs under set- and self-paced conditions over a 40-m distance. Self-paced conditions were performed at a self-selected pace for each activity. Set-paced conditions were performed within specific speed bands: walking, 1 to 2 $m{\cdot}s^{-1}$ (20- to 40-seconds); jogging, 3 to 4 $m{\cdot}s^{-1}$ (10- to 13-seconds); running, 5 to 6 $m{\cdot}s^{-1}$ (7- to 8-seconds). An RPE score was taken after each condition (set- and self-paced) using the Borg RPE 6-20 scale (340). All participants performed testing in their normal training attire which they wore for all sessions.

Participants in the concussed group were assessed at the following time points: return to run (RTR), RTP, 2-weeks post RTP, and 4-weeks post RTP. Number of days since concussion was calculated for each testing session for all concussed participants. For the control participants, they were matched with a concussed participant and provided with the same 'concussion date', as such, a 'day since concussion' variable was also calculated for controls based on the time between each testing session and the 'concussion date'. Participants within the control group followed the same time frames for testing wherever possible. Control players completed a post-concussion symptom scale (PCSS) prior to the first assessments. Concussed players completed a PCSS prior to each assessment.

### 5.3.3 Data processing

Following each testing session, the thoracic-mounted devices were connected to a computer running proprietary software (Sonra 2.0, Newry, Co. Down, N. Ireland). Individual drills were then created for each activity (walk, jog, run), repetition (i.e., walk 1), and condition (set- or self-paced). The start and end of each drill were determined by marking on the vertical accelerometer timeseries when movement began. Once all drills were created, the extended raw data (including drills and whole session) were exported to a comma-separated values (.csv) file. The raw data were exported as a continuous time series in 100 Hz files, with the 6-channel IMU data and instantaneous velocity provided. A separate .csv file containing the software derived metrics was also exported for each session. Raw data obtained from the Blue Trident IMUs were downloaded using proprietary software (Capture.U, Oxford, United Kingdom). Data for each condition (set- or self-paced) was exported as separate .csv files and saved in a separate password protected folder. Once the IMU data from each location were downloaded, all subsequent preprocessing and analysis of data were performed using the R Statistical Programming Language (R Foundation for Statistical Computing, Vienna, Austria) within the R Studio Integrated Development Environment (Posit, Boston, Massachusetts). Data from the GNSS and IMUs were processed using the same methods described in chapter four. Steps were established using a peak detection algorithm described in a previous study by using the vertical acceleration from each IMU (381).

### 5.3.4 Feature extraction

A range of spatiotemporal variables were calculated using each identified heel strike. Variables included: cadence (steps / time (s)); step length (difference in GNSS-derived distance (m) between consecutive steps); and the acceleration magnitude (*g*) at the point of each heal strike (antero-posterior, medio-lateral, vertical). Sample entropy, a non-linear analysis, was performed on mediolateral and vertical accelerations to assess gait regularity by analysing the entire time series (341). This was performed using the *sample_entropy* function in the *pracma* package, with the embedding dimensions set to two (*edim* = 2) and the tolerance threshold (*r*) set to 0.2 × the standard deviation of each time series. Additional variables; contact time (CT), flight time (FT), vertical GRF, and COM displacement were calculated by extracting vertical, mediolateral, and anterior-posterior variables from thoracic mounted devices using a validated algorithm (290) during jogging and running activities.

### 5.3.5 Statistical analyses

All statistical analyses were conducted in RStudio (version 2023.06.1+524, Posit, Boston, MA, USA) using the R programming language (version 4.2.3, R Foundation for Statistical Computing, Vienna, Austria) (refer to Appendix 6 for R Studio code used). To determine group differences across activities over time, Bayesian mixed effects models with a three-way group (concussed *vs.* control) × activity (walk *vs.* jog *vs.* run) × days since concussion interaction were used, with participant ID used as a random intercept term. There were no differences between set- and self-paced conditions, as such, the term for pace was removed from all models. Models were built using the *brm* function in the *brms* package (383). Bayesian models allow for a full posterior distribution to be estimated for the outcome variable based on the observed data and prior knowledge, capturing the uncertainty in model estimates appropriately. Model fitting and estimation of the posterior distribution is performed using Markov Chain Monte Carlo (MCMC) sampling. During initial model building, there were issues with convergence, as such model parameters were changed from the *brms* defaults. Each model used four separate Markov chains, with 4000 iterations within each chain, with 2000 iterations used for sampling, and a *treedepth* of 15. As such, 8000 draws were retained for estimating the posterior distribution within each model. Weak, informative population-level priors were provided for each model, where possible based on previous literature, when there was no obvious prior knowledge, noninformative priors were specified, these are highlighted in Table 5.1. As the effect of time was uncertain, non-informative flat priors were set for days since concussion within each model, so only the collected data were used to inform the posterior distribution. Priors were used to help inform robust estimates and model regularisation. Following running each model, convergence was checked via (1) inspecting the trace plots of the MCMC samples to ensure that the chains had mixed well, (2) ensuring the Gelman-Rubin (Rhat) statistic was close to 1; and (3) checking the effective sample size was large enough to ensure reliable parameters. A sensitivity analysis was performed by running the models for cadence and sample entropy with and without priors to ensure estimates were robust; with a posterior predictive check performed using the *pp_check* function to assess how well the posterior distribution fitted the observed data. All model estimates are reported with a 95% credible interval (CI). Hypothesis testing was performed using the *hypothesis* function within the *brms* package. Bayes factors (*BF*) were used to assess how well the posterior distributions supported the hypotheses that there would be more conservative, irregular gait within the concussed group that would improve over time (384). Interpretation of *BF* was determined via previously established thresholds (385); *BF* >1 and <1 show support for the alternative and null hypotheses, respectively with the specific thresholds shown in Table 5.2 Time to recovery was estimated from modelled data by providing the estimated number of days for the concussed group to cross zero. The full R code for each model is shown in Appendix 6.

Table 5.1 Priors used in each Bayesian mixed-effects model based on prior knowledge. Priors were stipulated for the differences between the concussed group and the control group at the initial assessment.

| Outcome variable | Term | Prior | Prior size | Prior knowledge |
|---|---|---|---|---|
| Step length | Group Concussed | Normal (-0.2, 0.1) | Weak informative | *d* = 0.7-1.0, Howell et al., 2018; Buckley et al., 2020 |
| Cadence | Group Concussed | Normal (-0.1, 0.03) | Weak informative | *d* = 0.5 - 1.1, Howell et al., 2018; Buckley et al., 2013 |
| Vertical acceleration | Group Concussed | Normal (-0.2, 0.05) | Weak informative | *d* = 0.3-0.9, Dubose et al., 2017 |
| Mediolateral acceleration | Group Concussed | Normal (0.05, 0.02) | Weak informative | *d* = 0.9, Catena et al., 2007 |
| Contact time | Group Concussed | Normal (0.1, 0.1) | Weak informative | *d* = 0.8-1.1, Martini et al., 2011; Buckley et al., 2016 |
| Flight time | Group Concussed | Normal (0, 0.3) | Non-informative | N/A |
| Vertical GRF | Group Concussed | Normal (0, 0.3) | Non-informative | N/A |
| COM displacement | Group Concussed | Normal (0, 0.3) | Non-informative | N/A |
| Sample entropy, vertical acceleration | Group Concussed | Normal (0.05, 0.02) | Weak informative | *d* = 0.4-0.6, Bonnette et al., 2020 |
| Sample entropy, mediolateral acceleration | Group Concussed | Normal (0.05, 0.02) | Weak informative | *d* = 0.4-0.6, Bonnette et al., 2020 |

All prior knowledge effect sizes (Cohen's d) were obtained from the initial assessment post-concussion in the reported studies. N/A denotes the non-informative, flat priors were used due to a lack of prior knowledge.

Table 5.2 Interpretation of Bayes factors (BF) to assess how well posterior distributions are supported by the hypotheses

| BF | Interpretation |
|---|---|
| >1 and <1 | Support for alternative and null hypotheses respectively |
| >150 | Very strong support of alternative hypotheses |
| >20 – 150 | Strong support of alternative hypotheses |
| 3 – 20 | Positive support of alternative hypotheses |
| 1, 0 | No support |
| >0.33 – 1 | Weak support for the null hypotheses |
| 0.05 – 0.33 | Positive support for the null hypotheses |
| 0.0067 – 0.05 | Strong support for the null hypotheses |
| <0.0067 | Very strong support for the null hypotheses |
| (385) | |

## 5.4 Results

For the concussed group, average PCSS scores were: RTR: 3.9 ± 6.9, RTP: 1.9 ± 3.9, two-weeks post RTP: 3.9 ± 6.7, and four-weeks post RTP: 1.3 ± 2.3. Of those eligible, a total of 43 sessions were completed by concussed participants (median = 3; range = 2 to 4; trials = 878) and 50 by control participants (median = 4; range = 2 to 4; trials = 1164). Concussed athletes missed testing sessions due to lower limb injury (n = 3; 2 at 2-weeks post RTP, 1 at 4-weeks post RTP), subsequent concussion (n = 1 at 4-weeks post RTP), and unavailability due to club commitments (n = 2; 1 at 2-weeks post RTP, 1 at 4-weeks post RTP). Control participants total PCSS score was 3.8 ± 9.9 at the first testing session. The characteristics of each activity performed during the initial testing session are displayed in Table 5.2 for control and concussed groups. Rating of perceived exertion (RPE) for each testing session is displayed in Table 5.3.

Table 5.3 Average, standard deviation, and effect size of differences between variables derived from lumbar- and thoracic-mounted IMUs for control and concussed groups during their initial testing session.

| | Thoracic spine | | | Lumbar spine | | |
|---|---|---|---|---|---|---|
| **Variable** | **Control group** | **Concussed group** | **Effect size (*d*) [95%CI]** | **Control group** | **Concussed group** | **Effect size (*d*) [95%CI]** |
| **Walk** | | | | | | |
| Speed *($m{\cdot}s^{-1}$)* | 1.3 ± 0.11 | 1.2 ± 0.12 | 0.99 [-0.07 to 2.03] | | | |
| Step length *(m)* | 0.73 ± 0.07 | 0.71 ± 0.04 | 0.34 [-0.67 to 1.35] | | | |
| Cadence *(steps/seconds)* | 1.81 ± 0.08 | 1.72 ± 0.09 | **1.35 [0.24 to 2.42]** | 1.82 ± 0.08 | 1.72 ± 0.09 | **1.28 [0.08 to 2.44]** |
| Acceleration vertical *($m{\cdot}s^{-2}$)* | 0.57 ± 0.12 | 0.41 ± 0.09 | **1.46 [0.34 to 2.54]** | 0.59 ± 0.15 | 0.53 ± 0.12 | 0.44 [-0.68 to 1.55] |
| Acceleration mediolateral *($m{\cdot}s^{-2}$)* | 0 ± 0.03 | 0 ± 0.02 | 0.03 [-0.97 to 1.03] | 0.02 ± 0.12 | 0.01 ± 0.15 | 0.17 [-0.94 to 1.27] |
| Sample entropy, vertical acceleration | 0.4 ± 0.06 | 0.39 ± 0.05 | 0.1 [-0.91 to 1.1] | 0.54 ± 0.1 | 0.54 ± 0.11 | -0.13 [-1.23 to 0.97] |
| Sample entropy, mediolateral acceleration | 0.5 ± 0.11 | 0.47 ± 0.11 | 0.23 [-0.78 to 1.23] | 0.62 ± 0.11 | 0.7 ± 0.07 | -0.8 [-1.92 to 0.35] |
| **Jog** | | | | | | |
| Speed *($m{\cdot}s^{-1}$)* | 3.27 ± 0.61 | 3.02 ± 0.63 | 0.48 [-0.55 to 1.49] | | | |
| Step length *(m)* | 1.22 ± 0.2 | 1.13 ± 0.21 | 0.52 [-0.51 to 1.53] | | | |
| Cadence *(steps/seconds)* | 2.77 ± 0.14 | 2.75 ± 0.1 | 0.16 [-0.84 to 1.17] | 2.77 ± 0.12 | 2.73 ± 0.1 | 0.31 [-0.8 to 1.41] |
| Acceleration vertical *($m{\cdot}s^{-2}$)* | 2.83 ± 0.66 | 2.56 ± 0.51 | 0.44 [-0.58 to 1.45] | 2.7 ± 0.4 | 2.98 ± 1.14 | -0.21 [-1.31 to 0.9] |
| Acceleration mediolateral *($m{\cdot}s^{-2}$)* | -0.02 ± 0.24 | 0.04 ± 0.22 | -0.25 [-1.25 to 0.76] | 0.25 ± 0.42 | 0.58 ± 0.57 | -0.9 [-2.03 to 0.26] |
| Sample entropy, vertical acceleration | 0.35 ± 0.06 | 0.4 ± 0.07 | -0.84 [-1.87 to 0.21] | 0.41 ± 0.09 | 0.46 ± 0.16 | -0.53 [-1.64 to 0.6] |
| Sample entropy, mediolateral acceleration | 0.64 ± 0.11 | 0.68 ± 0.07 | -0.4 [-1.41 to 0.61] | 0.69 ± 0.14 | 0.68 ± 0.13 | 0.04 [-1.06 to 1.14] |
| Contact time *(s)* | 0.25 ± 0.02 | 0.24 ± 0.02 | 0.62 [-0.42 to 1.64] | | | |
| Flight time *(s)* | 0.12 ± 0.02 | 0.13 ± 0.02 | -0.77 [-1.81 to 0.28] | | | |
| Vertical GRF *(N)* | 2.04 ± 0.29 | 2.3 ± 0.29 | -0.9 [-1.94 to 0.16] | | | |
| COM displacement *(m)* | 0.08 ± 0.01 | 0.07 ± 0.01 | 0.63 [-0.41 to 1.66] | | | |
| Vertical stiffness (Kvert) *($kN{\cdot}m^{-1}$)* | 30.67 ± 9.49 | 37.66 ± 9.62 | -0.99 [-2.04 to 0.08] | | | |

Table 5.3 continued

| | | | | | | |
|---|---|---|---|---|---|---|
| | | | **Run** | | | |
| Speed $(m{\cdot}s^{-1})$ | 4.64 ± 0.49 | 4.95 ± 0.56 | -0.72 [-1.74 to 0.32] | | | |
| Step length *(m)* | 1.58 ± 0.12 | 1.58 ± 0.12 | 0.01 [-1 to 1.01] | | | |
| Cadence *(steps/seconds)* | 3.07 ± 0.19 | 3.28 ± 0.26 | -1.03 [-2.08 to 0.04] | 3.05 ± 0.18 | 3.25 ± 0.18 | **-1.35 [-2.52 to -0.14]** |
| Acceleration vertical $(m{\cdot}s^{-2})$ | 3.29 ± 0.71 | 3.24 ± 0.59 | 0.08 [-0.93 to 1.08] | 2.92 ± 0.35 | 3.65 ± 0.79 | **-1.43 [-2.62 to -0.21]** |
| Acceleration mediolateral $(m{\cdot}s^{-2})$ | 0 ± 0.28 | 0.07 ± 0.27 | -0.25 [-1.25 to 0.76] | 0.22 ± 0.45 | 0.4 ± 0.78 | -0.36 [-1.46 to 0.76] |
| Sample entropy, vertical acceleration | 0.49 ± 0.11 | 0.62 ± 0.09 | **-1.25 [-2.32 to -0.16]** | 0.53 ± 0.13 | 0.65 ± 0.11 | -1.06 [-2.21 to 0.11] |
| Sample entropy, mediolateral acceleration | 0.75 ± 0.12 | 0.89 ± 0.06 | **-1.29 [-2.36 to -0.19]** | 0.76 ± 0.13 | 0.83 ± 0.11 | -0.53 [-1.64 to 0.6] |
| Contact time *(s)* | 0.24 ± 0.03 | 0.22 ± 0.02 | 0.76 [-0.29 to 1.79] | | | |
| Flight time *(s)* | 0.09 ± 0.01 | 0.09 ± 0.01 | 0 [-1.02 to 0.97] | | | |
| Vertical GRF *(N)* | 1.95 ± 0.29 | 2.11 ± 0.23 | -0.59 [-1.61 to 0.45] | | | |
| COM displacement *(m)* | 0.07 ± 0.02 | 0.06 ± 0.01 | 0.76 [-0.29 to 1.79] | | | |
| Vertical stiffness (Kvert) $(kN{\cdot}m^{-1})$ | 39.72 ± 56.95 | 39.19 ± 11.51 | 0.02 [-0.99 to 1.03] | | | |

Bolded text indicates statistically significant difference between control and concussed groups with the confidence interval not containing zero. Positive effect size represents a higher score for the control group, whereas a negative effect size represents a higher score for the concussed group

GRF, vertical ground reaction force; Kvert, vertical stiffness; COM, centre of mass; $m{\cdot}s^{-1}$, meters per second; $m{\cdot}s^{-2}$, meters per second squared; N, Newtons, $kN{\cdot}m^{-1}$, kilonewton per meter

Table 5.4 Average and standard deviation of Rating of Perceived Exertion (RPE) scores for concussed and control participants across testing sessions

| Group | Session 1 | Session 2 | Session 3 | Session 4 |
|---|---|---|---|---|
| Control | 10 ± 2.2 | 9.7 ± 2.1 | 9.8 ± 2.1 | 10.2 ± 2.4 |
| Concussed | 10.5 ± 1.6 | 10 ± 1.8 | 10.1 ± 1.9 | 9.5 ± 2.0 |

Session RPE scores with set- and self-pace conditions combined

### 5.4.1 Linear measures

5.4.1.1 Step length

There was *weak* and uncertain evidence to support shorter step lengths when walking in the concussed group (estimate [95%CI] = -0.01 [-0.07 to 0.05], *BF* = 2.02), which did not support a clear improvement over time (*BF* = 0.45). This was not the case for higher velocities though, with *strong* and *very strong* evidence to support the null hypothesis, with longer step lengths amongst the concussed group for jog (estimate [95%CI] = 0.05 [0.01 to 0.09], *BF* = 0.01) and run (estimate [95%CI] = 0.06 [0.02 to 0.102], *BF* = 0) respectively, both of which did not change over time (*BF* = 0).

5.4.1.2 Cadence

There was *strong* evidence for the concussed participants having slower cadence (Figure 5.1), when walking at both IMU sites compared to the control group (thoracic: estimate [95%CI] = -0.05 [-0.1 to 0], *BF* = 31; lumbar: estimate [95%CI] = -0.05 [-0.1 to 0], *BF* = 37.3). The slower walking cadence persisted across the different time points for the concussed group, with no evidence to support an increase in cadence over time (thoracic *BF* = 0; lumbar *BF* = 0.04). In contrast, there was little difference between the groups with respect to cadence while jogging (thoracic: estimate [95%CI] = 0.01 [-0.028 to 0.047]; lumbar: estimate [95%CI] = -0.03 [-0.07 to 0.01]), but the concussed group displayed higher cadence during the running trials (thoracic: estimate [95%CI] = 0.11 [0.07 to 0.15]; lumbar: estimate [95%CI] = 0.05 [0.01 to 0.096]).

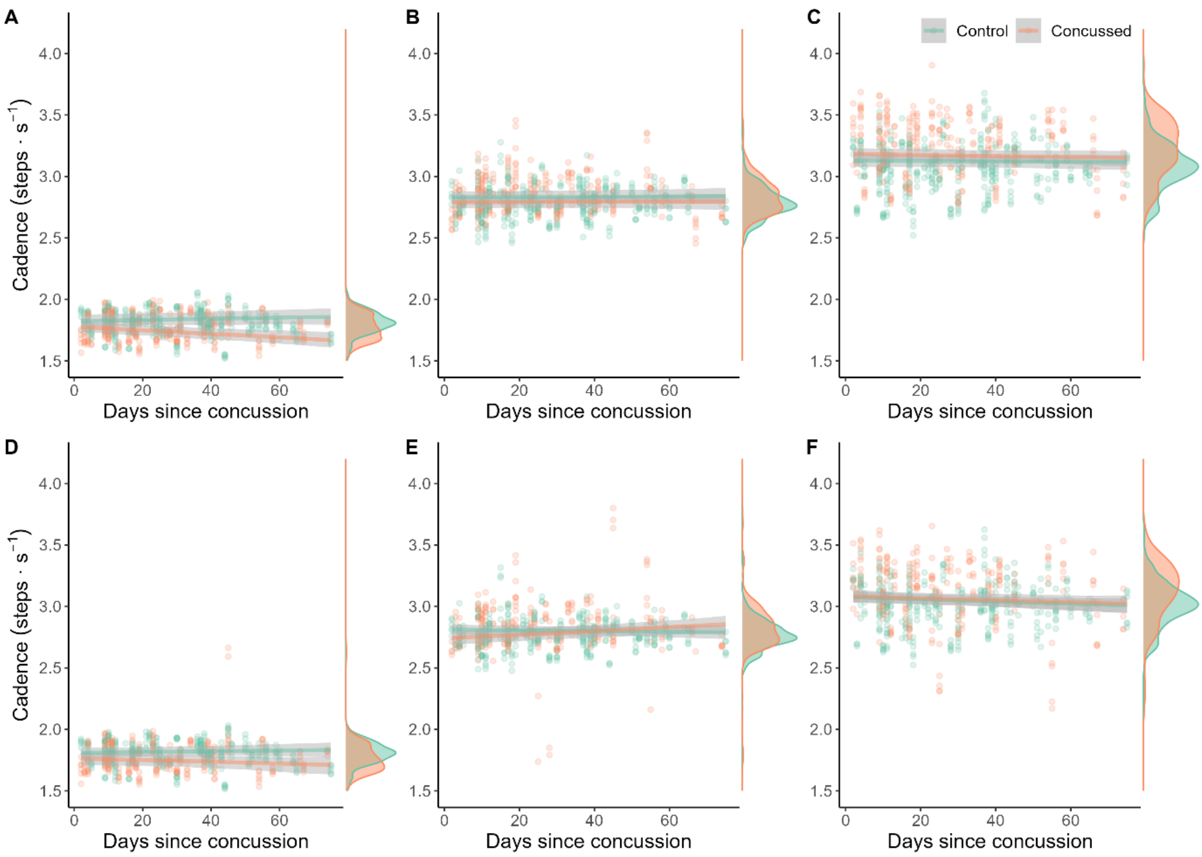


Figure 5.1 Differences between groups for cadence at the thoracic spine (A-C) and lumbar spine (D-F) during walking (A, D), jogging (B, E) and running (C, F) activities. Data are the fitted line and 95% credible interval taken from 200 draws per sample from the posterior distribution of the Bayesian mixed effects models; raw data are shown as the points along with their distribution in the margins.

#### 5.4.1.3 Acceleration, vertical

There was *very strong* evidence to support reduced vertical acceleration when walking for thoracic- and lumbar-mounted IMUs (thoracic: estimate [95%CI] = -0.25 [-0.41 to -0.09], *BF* = 1999; lumbar: estimate [95%CI] = -0.22 [-0.39 to -0.06], *BF* = 215). Vertical accelerations increased in the concussed group when walking as time since concussion increased at the thoracic spine, (*BF* = 112), with an estimated recovery by 58 days post-concussion (95%CI = 25 to 90 days). There was no improvement at the lumbar spine (*BF* = 0.32). The concussed group showed greater vertical acceleration when jogging and running (thoracic: jog: estimate [95%CI] = 0.19 [0.05 to 0.32], *BF* = 0, run: Estimate [95%CI] = 0.203 [0.06 to 0.34], *BF* = 0; lumbar: jog: estimate [95%CI] = 0.15 [-0.01 to 0.3], *BF* = 0.03, run: estimate [95%CI] = 0.46 [0.31 to 0.61], *BF* = 0), which remained consistent over time.

#### 5.4.1.4 Acceleration, mediolateral

Concussed participants displayed increased mediolateral accelerations, with *strong* and *very strong* evidence to support this when walking (thoracic: estimate [95%CI] = 0.04 [0.00 to 0.07], *BF* = 51.29; lumbar: estimate [95%CI] = 0.06 [0.02 to 0.09], *BF* = 887.89), jogging (thoracic: estimate [95%CI] = 0.104 [0.07 to 0.14], *BF* = 8000; lumbar: estimate [95%CI] = 0.101 [0.06 to 0.14], *BF* = 8000) and running (thoracic: estimate [95%CI] = 0.08 [0.05 to 0.12], *BF* = 8000; lumbar: estimate [95%CI] = 0.102 [0.06 to 0.14], *BF* = 8000). Over time, there were no clear improvements in the concussed group at the thoracic site when walking (*BF* = 0.35); there was a trend for improvements during jogging (*BF* = 48.38) and running (*BF* = 3.85), but the credible intervals spanning zero highlight uncertainty over time. At the lumbar site, there was *very strong* evidence for improvements when walking (*BF* = 8000; recovery time = 13; 4 to 22 days), jogging (*BF* = 8000; recovery time = 14, 9 to 20 days) and running (*BF* = 8000; recovery time = 13, 9 to 19 days).

#### 5.4.1.5 Contact time

Concussed athletes tended to display shorter contact times, with *positive* evidence to support the null hypothesis during the jog (estimate [95%CI] = -0.01 [-0.02 to 0.001], *BF* = 0.06) and run (estimate [95%CI] = -0.003 [-0.01 to 0.004], *BF* = 0.27) activities. Over time, the concussed group showed no changes in contact time for jog (*BF* = 0) and run (*BF* = 0.2) activities.

#### 5.4.1.6 Flight time

There was no evidence to support shorter flight times when jogging in the concussed group, with *positive* evidence to support the null hypothesis when jogging (estimate [95%CI] = 0.01 [-0.003 to 0.01], *BF* = 0.11). However, there was *weak* evidence that concussed participants had slightly shorter flight times than controls while running (estimate [95%CI] = -0.002 [-0.01 to 0.004], *BF* = 2.98). Neither activity showed improvement over time (jog: *BF* = 0; run: *BF* = 0.03).

5.4.1.7 Vertical GRF

There was *positive* evidence to support lower vertical GRF when running in the concussed group (estimate[95%CI] = -0.04 [-0.06 to 0.03], *BF* = 5.02) but no difference when jogging (estimate [95%CI] = 0.001 [-0.23 to 0.23], *BF* = 0.97). Over time, there were no improvements in vertical GRF for jog (*BF* = 0) or run (*BF* = 0.08) activities.

5.4.1.8 COM displacement

There was less COM displacement in concussed athletes when jogging and running, with *positive* (estimate [95%CI] = -0.01 [-0.01 to 0.001], *BF* = 19.78) and *weak* evidence (estimate [95%CI] = -0.001 [-0.01 to 0.003], *BF* = 2.7), respectively. Over time, COM displacement improved (increased) faster when jogging (*BF* = 2665.67) and slowest when running (*BF* = 5.58).

### 5.4.2 Non-linear analyses

5.4.2.1 Sample entropy

There was evidence to support increased sample entropy in the concussed group in both vertical (Figure 5.2) and mediolateral accelerations (Figure 5.3). This was apparent across all activities for vertical accelerations, with greater differences seen as velocity increased for both the thoracic (walk: estimate [95%CI] = 0.018 [-0.01 to 0.05], *BF* = 9.26, jog: estimate [95%CI] = 0.034 [0.01 to 0.06], *BF* = 332.33, run: estimate [95%CI] = 0.12 [0.09 to 0.14], *BF* = 8000) and lumbar sites (walk: estimate [95%CI] = 0.03 [0 to 0.06], *BF* = 38.22, jog: estimate [95%CI] = 0.04 [0.01 to 0.07], *BF* = 257.06, run: estimate [95%CI] = 0.115 [0.09 to 0.15], *BF* = 8000). Whilst the largest differences were seen when running, it was the only activity that showed clear improvements over time at both thoracic (*BF* = 8000) and lumbar (*BF* = 295.3) locations. Albeit small, the concussed group still had elevated sample entropy at 75 days post-concussion. Improvements in walking (*BF* = 12.91) and jogging (*BF* = 1.12) were only seen at the lumbar spine; with the thoracic spine showing no improvements (walk *BF* = 0.07; jog *BF* = 0).

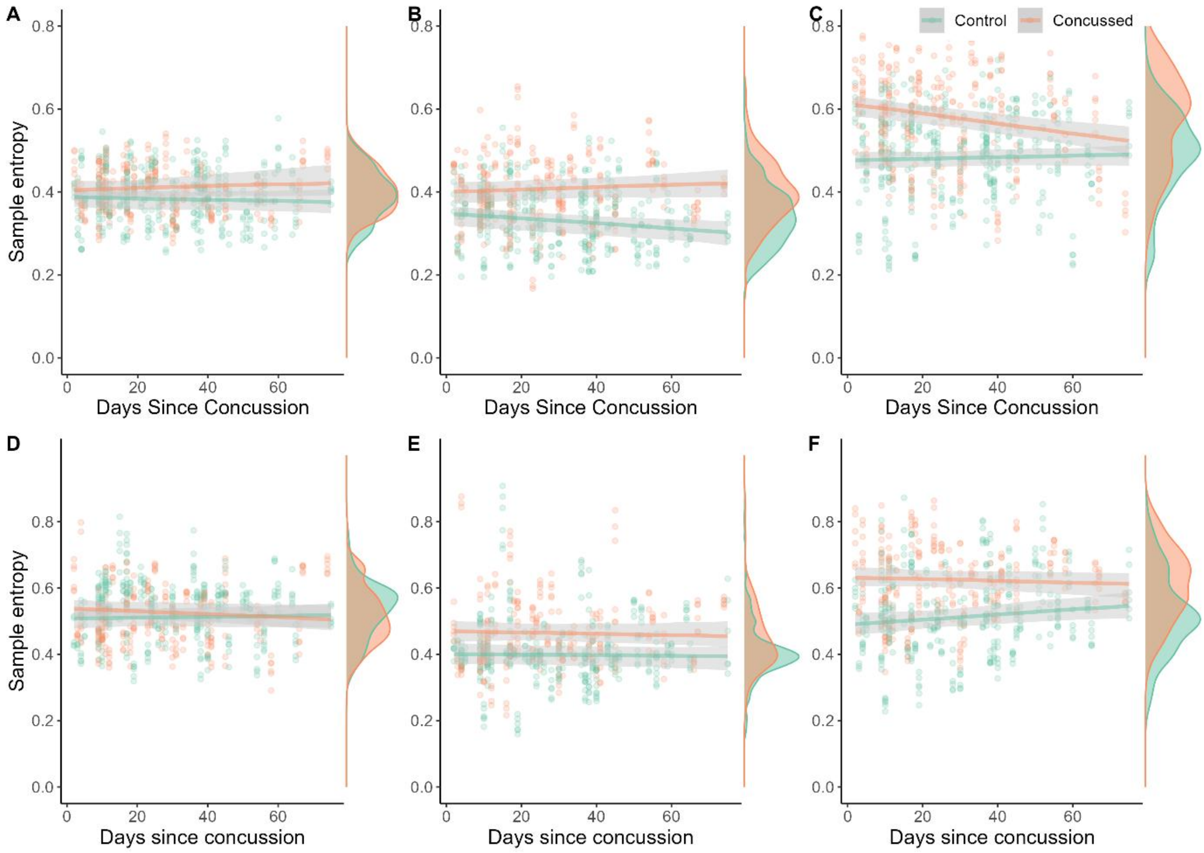


Figure 5.2 Differences between groups for vertical acceleration sample entropy at the thoracic spine (A-C) and lumbar spine (D-F) during walking (A, D), jogging (B, E) and running (C, F) activities. Data are the fitted line and 95% credible interval taken from 200 draws per sample from the posterior distribution of the Bayesian mixed effects models; raw data are shown as the points along with their distribution in the margins.

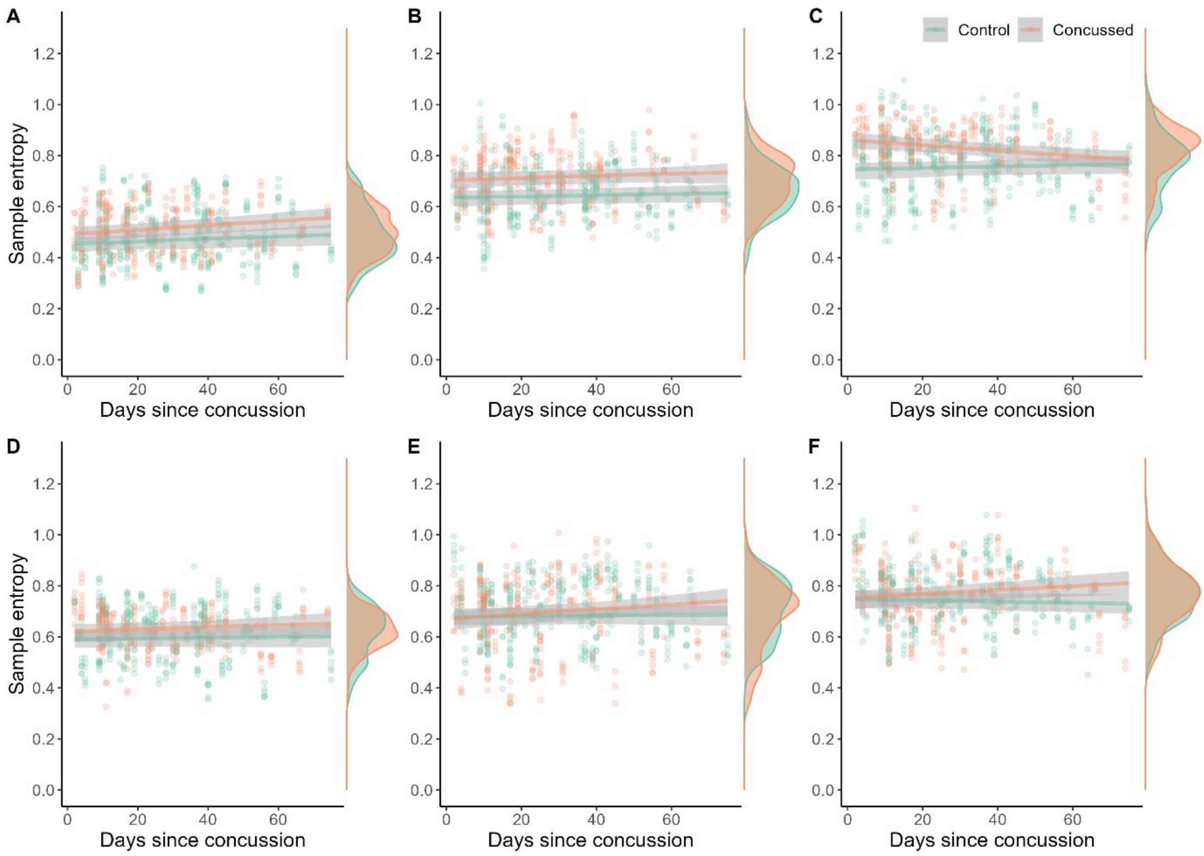


Figure 5.3 Differences between groups for mediolateral acceleration sample entropy at the thoracic spine (A-C) and lumbar spine (D-F) during walking (A, D), jogging (B, E) and running (C, F) activities. Data are the fitted line and 95% credible interval taken from 200 draws per sample from the posterior distribution of the Bayesian mixed effects models; raw data are shown as the points along with their distribution in the margins.

For mediolateral accelerations there was *strong* and *very strong* evidence to support increased entropy amongst the concussed group for each activity when derived from thoracic-mounted IMUs (walk: estimate [95%CI] = 0.04 [0.003 to 0.07], *BF* = 61.5, jog: estimate [95%CI] = 0.03 [0.001 to 0.06], *BF* = 46.9, run: 0.08 [0.05 to 0.11], *BF* = 8000) and *strong* evidence to support this when walking for lumbar-mounted IMUs (estimate [95%CI] = 0.03 [-0.002 to 0.07], *BF* = 29.89). Jog and run activities displayed little difference in mediolateral entropy for the lumbar spine (jog: estimate [95%CI] = -0.04 [-0.07 to -0.002], *BF* = 0.02, run: -0.03 [-0.07 to 0.01], *BF* = 0.07). Over time there was little change in entropy when derived from the lumbar spine (walk *BF* = 0.33, jog *BF* = 0.3, run *BF* = 0.01) and at the thoracic spine when walking (*BF* = 0.19) and jogging (*BF* = 0.52). However, there was *very strong* evidence to support improved mediolateral entropy scores over time at the thoracic spine when running (*BF* = 3999, recovery time = 77, 46 to 106 days).

Table 5.5 Bayesian model outputs and hypothesis testing for outcome measures derived from lumbar- and thoracic-mounted IMUs during walk, jog, and run activities

| **Term** | **Estimate [95% CI]** | **Bulk ESS** | **Tail ESS** | **Estimate [95% CI]** | **Bulk ESS** | **Tail ESS** | **Hypothesis test for concussed group** | **BF** | **Posterior Probability** | **BF** | **Posterior Probability** |
|---|---|---|---|---|---|---|---|---|---|---|---|
| **Step length (SL) (m)** | Thoracic | | | | | | | Thoracic | | Lumbar | |
| Control, Walk (intercept) | 0.738 [0.697 to 0.781] | 828 | 1893 | | | | | | | | |
| Concussed x Walk | -0.014 [-0.073 to 0.045] | 1002 | 1862 | | | | ↓ SL | 2.02 | 0.67 | | |
| Concussed x Jog | 0.05 [0.008 to 0.092] | 3035 | 4322 | | | | ↓ SL | 0.01 | 0.01 | | |
| Concussed x Run | 0.059 [0.016 to 0.102] | 2737 | 4506 | | | | ↓ SL | 0 | 0 | | |
| Concussed x Walk x Days Since Concussion | 0 [-0.001 to 0.001] | 4161 | 5435 | | | | ↑ SL over time | 0.45 | 0.31 | | |
| Concussed x Jog x Days Since Concussion | -0.001 [-0.002 to 0] | 3732 | 5398 | | | | ↑ SL over time | 0 | 0 | | |
| Concussed x Run x Days Since Concussion | -0.001 [-0.003 to 0] | 3544 | 5375 | | | | ↑ SL over time | 0 | 0 | | |
| **Cadence (steps·s$^{-1}$)** | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
| Control, Walk (intercept) | 1.824 [1.774 to 1.875] | 886 | 1805 | 1.812 [1.765 to 1.862] | 1384 | 2461 | | | | | |
| Concussed x Walk | -0.047 [-0.099 to 0.003] | 1448 | 2689 | -0.048 [-0.098 to 0] | 1975 | 2920 | ↓ cadence | 31 | 0.97 | 37.28 | 0.97 |
| Concussed x Jog | 0.009 [-0.028 to 0.047] | 3291 | 4138 | -0.028 [-0.068 to 0.013] | 4934 | 5406 | ↓ cadence | 0.46 | 0.31 | 9.65 | 0.91 |
| Concussed x Run | 0.109 [0.072 to 0.145] | 3632 | 4746 | 0.054 [0.013 to 0.096] | 5043 | 5330 | ↓ cadence | 0 | 0 | 0 | 0 |
| Concussed x Walk x Days Since Concussion | -0.002 [-0.003 to -0.001] | 4156 | 6689 | -0.001 [-0.002 to 0] | 5275 | 6402 | ↑ cadence over time | 0 | 0 | 0.04 | 0.04 |
| Concussed x Jog x Days Since Concussion | 0.002 [0.001 to 0.003] | 4705 | 6095 | 0.003 [0.001 to 0.004] | 6137 | 6770 | ↑ cadence over time | 1.1 | 0.52 | 499 | 1 |
| Concussed x Run x Days Since Concussion | 0.001 [0 to 0.003] | 4707 | 6555 | 0.001 [0 to 0.002] | 6636 | 6608 | ↑ cadence over time | 0.27 | 0.22 | 1.18 | 0.54 |
| **Vertical acceleration (g)** | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
| Control, Walk (intercept) | 0.698 [0.544 to 0.847] | 1212 | 2072 | 0.719 [0.533 to 0.903] | 1047 | 1868 | | | | | |
| Concussed x Walk | -0.247 [-0.41 to -0.092] | 1829 | 2953 | -0.222 [-0.386 to -0.057] | 2002 | 3611 | ↓ vertical acceleration | 1999 | 1 | 215.22 | 1 |
| Concussed x Jog | 0.185 [0.053 to 0.317] | 3198 | 4557 | 0.149 [-0.008 to 0.3] | 3056 | 4502 | ↓ vertical acceleration | 0 | 0 | 0.03 | 0.03 |
| Concussed x Run | 0.203 [0.068 to 0.336] | 2984 | 4266 | 0.458 [0.312 to 0.611] | 2948 | 3968 | ↓ vertical acceleration | 0 | 0 | 0 | 0 |
| Concussed x Walk x Days Since Concussion | 0.004 [0.001 to 0.007] | 3658 | 5204 | -0.001 [-0.005 to 0.002] | 3633 | 4790 | ↑ vertical acceleration over time | 111.68 | 0.99 | 0.32 | 0.24 |
| Concussed x Jog x Days Since Concussion | -0.008 [-0.013 to -0.004] | 3447 | 5099 | 0.002 [-0.003 to 0.007] | 3432 | 4905 | ↑ vertical acceleration over time | 0 | 0 | 2.06 | 0.67 |
| Concussed x Run x Days Since Concussion | -0.005 [-0.009 to -0.001] | 3434 | 5256 | -0.005 [-0.01 to 0] | 3338 | 4369 | ↑ vertical acceleration over time | 0.42 | 0.29 | 0 | 0 |

Table 5.5 continued

| Mediolateral (ML) acceleration (g) | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Control, Walk (intercept) | -0.037 [-0.088 to 0.014] | 1023 | 2005 | 0.047 [-0.054 to 0.146] | 1316 | 2617 | | | | | |
| Concussed x Walk | 0.038 [0.003 to 0.074] | 2349 | 3762 | 0.055 [0.017 to 0.092] | 6949 | 5598 | ↑ ML acceleration | 51.29 | 0.98 | 887.89 | 1 |
| Concussed x Jog | 0.104 [0.072 to 0.136] | 4072 | 4701 | 0.101 [0.063 to 0.138] | 9943 | 5404 | ↑ ML acceleration | 8000 | 1 | 8000 | 1 |
| Concussed x Run | 0.083 [0.051 to 0.115] | 4171 | 5351 | 0.102 [0.064 to 0.14] | 8062 | 5462 | ↑ ML acceleration | 8000 | 1 | 8000 | 1 |
| Concussed x Walk x Days Since Concussion | 0 [-0.001 to 0.001] | 5957 | 6656 | -0.005 [-0.008 to -0.002] | 6519 | 6145 | ↓ ML acceleration over time | 0.35 | 0.26 | 8000 | 1 |
| Concussed x Jog x Days Since Concussion | -0.002 [-0.003 to 0] | 6435 | 6499 | -0.001 [-0.004 to 0.001] | 10377 | 6708 | ↓ ML acceleration over time | 48.38 | 0.98 | 8000 | 1 |
| Concussed x Run x Days Since Concussion | -0.001 [-0.002 to 0] | 6835 | 6662 | 0 [-0.003 to 0.003] | 11458 | 7013 | ↓ ML acceleration over time | 3.85 | 0.79 | 3999 | 1 |

| Contact time (CT) (s) | Thoracic | | | | | | | Thoracic | | Lumbar | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Control, Jog (intercept) | 0.294 [0.286 to 0.303] | 1690 | 2835 | | | | | | | | |
| Concussed x Jog | -0.008 [-0.016 to 0.001] | 1064 | 1698 | | | | ↑ CT | 0.06 | 0.05 | | |
| Concussed x Run | -0.003 [-0.009 to 0.004] | 5006 | 5519 | | | | ↑ CT | 0.27 | 0.21 | | |
| Concussed x Jog x Days Since Concussion | 0 [0 to 0] | 6870 | 6715 | | | | ↓ CT over time | 0 | 0 | | |
| Concussed x Run x Days Since Concussion | 0 [0 to 0] | 5799 | 5690 | | | | ↓ CT over time | 0.2 | 0.17 | | |

| Flight time (FT) (s) | Thoracic | | | | | | | Thoracic | | Lumbar | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Control, Jog (intercept) | 0.153 [0.146 to 0.16] | 3448 | 5214 | | | | | | | | |
| Concussed x Jog | 0.005 [-0.003 to 0.012] | 1392 | 2659 | | | | ↓ FT | 0.11 | 0.1 | | |
| Concussed x Run | -0.002 [-0.008 to 0.004] | 8080 | 5896 | | | | ↓ FT | 2.98 | 0.75 | | |
| Concussed x Jog x Days Since Concussion | 0 [0 to 0] | 7885 | 6516 | | | | ↑ FT over time | 0 | 0 | | |
| Concussed x Run x Days Since Concussion | 0 [0 to 0] | 7992 | 6208 | | | | ↑ FT over time | 0.03 | 0.03 | | |

| Vertical GRF (kN) | Thoracic | | | | | | | Thoracic | | Lumbar | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Control, Jog (intercept) | 2.181 [2.015 to 2.349] | 1115 | 1862 | | | | | | | | |
| Concussed x Jog | 0.001 [-0.234 to 0.229] | 997 | 1948 | | | | ↓ vGRF | 0.97 | 0.49 | | |
| Concussed x Run | -0.043 [-0.057 to -0.028] | 3663 | 3587 | | | | ↓ vGRF | 5.02 | 0.83 | | |
| Concussed x Jog x Days Since Concussion | -0.024 [-0.072 to 0.025] | 3464 | 3698 | | | | ↑ vGRF over time | 0 | 0 | | |
| Concussed x Run x Days Since Concussion | -0.002 [-0.003 to -0.001] | 5643 | 6308 | | | | ↑ vGRF over time | 0.08 | 0.07 | | |

Table 5.5 continued

| COM displacement (cm) | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| Control, Jog (intercept) | 0.104 [0.098 to 0.109] | 1419 | 3190 | | | | | | | | |
| Concussed x Jog | -0.005 [-0.01 to 0.001] | 924 | 1725 | | | | ↓ COM displacement | 19.78 | 0.95 | | |
| Concussed x Run | -0.008 [-0.009 to -0.006] | 7980 | 6597 | | | | ↓ COM displacement | 2.7 | 0.73 | | |
| Concussed x Jog x Days Since Concussion | -0.001 [-0.005 to 0.003] | 7340 | 6021 | | | | ↑ COM displacement over time | 2665.67 | 1 | | |
| Concussed x Run x Days Since Concussion | 0 [0 to 0] | 7245 | 6712 | | | | ↑ COM displacement over time | 5.58 | 0.85 | | |
| **Sample entropy, vertical acceleration** | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
| Control, Walk (intercept) | 0.386 [0.363 to 0.408] | 1084 | 2002 | 0.508 [0.478 to 0.536] | 877 | 1931 | | | | | |
| Concussed x Walk | 0.018 [-0.008 to 0.047] | 1419 | 2608 | 0.031 [0 to 0.064] | 1189 | 2372 | ↑ vertical entropy | 9.26 | 0.9 | 38.22 | 0.97 |
| Concussed x Jog | 0.034 [0.01 to 0.059] | 4124 | 4409 | 0.039 [0.01 to 0.068] | 2355 | 3984 | ↑ vertical entropy | 332.33 | 1 | 257.06 | 1 |
| Concussed x Run | 0.118 [0.092 to 0.142] | 4142 | 5209 | 0.115 [0.085 to 0.145] | 2245 | 3751 | ↑ vertical entropy | 8000 | 1 | 8000 | 1 |
| Concussed x Walk x Days Since Concussion | 0 [0 to 0.001] | 5193 | 6456 | -0.001 [-0.001 to 0] | 3150 | 5607 | ↓ vertical entropy over time | 0.07 | 0.07 | 12.91 | 0.93 |
| Concussed x Jog x Days Since Concussion | 0 [0 to 0.001] | 4973 | 5736 | 0 [0 to 0.001] | 3209 | 4850 | ↓ vertical entropy over time | 0 | 0 | 1.12 | 0.53 |
| Concussed x Run x Days Since Concussion | -0.002 [-0.003 to -0.001] | 4857 | 6095 | -0.001 [-0.001 to 0] | 3078 | 5100 | ↓ vertical entropy over time | 8000 | 1 | 295.3 | 1 |
| **Sample entropy, mediolateral (ML) acceleration** | Thoracic | | | Lumbar | | | | Thoracic | | Lumbar | |
| Control, Walk (intercept) | 0.455 [0.423 to 0.485] | 844 | 1583 | 0.588 [0.553 to 0.621] | 792 | 1704 | | | | | |
| Concussed x Walk | 0.035 [0.003 to 0.066] | 1427 | 2730 | 0.031 [-0.002 to 0.065] | 1402 | 2293 | ↑ ML entropy | 61.5 | 0.98 | 29.89 | 0.97 |
| Concussed x Jog | 0.032 [0.001 to 0.063] | 2282 | 3640 | -0.038 [-0.074 to -0.002] | 2451 | 3739 | ↑ ML entropy | 46.9 | 0.98 | 0.02 | 0.02 |
| Concussed x Run | 0.08 [0.049 to 0.111] | 2311 | 3414 | -0.028 [-0.065 to 0.009] | 2550 | 4071 | ↑ ML entropy | 8000 | 1 | 0.07 | 0.06 |
| Concussed x Walk x Days Since Concussion | 0 [0 to 0.001] | 2937 | 4987 | 0 [-0.001 to 0.001] | 2712 | 4500 | ↓ ML entropy over time | 0.19 | 0.16 | 0.33 | 0.25 |
| Concussed x Jog x Days Since Concussion | 0 [-0.001 to 0.001] | 2862 | 4766 | 0.001 [-0.001 to 0.002] | 2967 | 4698 | ↓ ML entropy over time | 0.52 | 0.34 | 0.03 | 0.03 |
| Concussed x Run x Days Since Concussion | -0.002 [-0.003 to -0.001] | 2821 | 4234 | 0.001 [0 to 0.002] | 3331 | 4708 | ↓ ML entropy over time | 3999 | 1 | 0.01 | 0.01 |

Abbreviations. SL, step length; CT, contact time; FT, flight time; vGRF, vertical ground reaction force; COM, centre of mass; ML, mediolateral; CI, confidence interval; BF, Bayes factor; ↑, increase; ↓, decrease.

## 5.5 Discussion

The primary aim of this study was to assess changes in spatiotemporal and nonlinear gait metrics using lumbar- and thoracic-mounted IMUs during walking, jogging, and running for up to four weeks post RTP from a concussion. Our findings suggest that variations in gait strategies exist beyond normal RTP timelines and may be present for more than 60-days post-concussion for selected outcome measures, some of these differences are velocity dependent. Concussed athletes demonstrated alterations in gait strategies with no between group differences in RPE scores. Concussed athletes were mostly asymptomatic at the time of testing and cleared for RTR activities at greater intensities than those performed in the study. This may explain why no difference in RPE score was seen between groups. The concussed group showed fewer regular movements in the mediolateral and vertical direction across all velocities (denoted by the increased sample entropy values), despite reduced COM displacement. When walking, the concussed group displayed lower cadence, that did not improve over time; this was not seen at higher velocities. Lumbar- and thoracic-mounted devices showed similar findings for linear (cadence, step time, and vertical and mediolateral acceleration profiles) and non-linear measure of gait. Whilst movement signals from thoracic-mounted devices may be blunted due to their distance up the kinetic chain, practitioners can use these devices to assess concussion-induced changes in gait during constant velocity movements that may persist beyond RTP. These devices could be used to objectively monitor changes in the field to inform individualised RTP strategies; movement assessment post-concussion should be performed across a range of velocities.

This study demonstrated that increasing movement velocity can impact selected spatiotemporal gait parameters. Additionally, changes in movement strategies seem to exist without an increased perception of effort in the concussed athletes. Until now, gait analysis has only assessed walking, describing concussed athletes as having a more conservative gait (266, 267, 361). This is characterised by slower walking velocities, increased stride times, greater irregularity in frontal and sagittal planes, and reductions in COM displacement from the base of support (BoS), perhaps due to compromised dynamic balance (131, 361, 386, 387). Our findings support this, with concussed athletes demonstrating a slower cadence during walking trials compared to controls. However, measures of cadence and vertical acceleration during jogging and running trials indicated concussed athletes were moving at equal or greater gait velocities when compared to controls. Gait is largely controlled via sensory feedback provided by vision, vestibular and proprioceptors that influence movement patterns (e.g. to accommodate for changes in the walking surface) (109). During walking, there is a greater reliance on attentional demand and sensory feedback in comparison to jog and run tasks which are more rhythmically constrained (388). In addition, performing these tasks on natural grassed surfaces may introduce greater variability and more reliance on a highly attuned sensory feedback system than predictable artificial surfaces (e.g. indoor court or laboratory). With regards to our findings, it is plausible that the activity of walking allows more time to process sensory information, therefore allowing more time to modify movement strategy. Following concussion, athletes may experience hypoexcitability of the cortical pathways or increased intra-cortical inhibition. As a result, disrupted communications between the pre-motor area, supplementary motor area, and the primary motor cortex may impact planning, coordination, and execution of movements (123, 131). Specifically, we may see dampening of neural signals resulting in slower, less efficient motor commands. Gait adaptability may also be affected as the neural system may be less flexible in accommodating for changes in movement patterns that are often required when walking at slower velocities (388). In the context of cadence, concussed athletes may take less steps per minute (lower cadence) during walking activities due to the combination of increased sensory feedback and a reduced ability to process and execute movement strategies in a timely manner. Whereas, during jog and run activities, a higher cadence (taking more steps) in conjunction with more rhythmic movements may provide greater stability over increasing velocities, accommodating for a less coordinated system.

The increased mediolateral accelerations and irregularity in the concussed group, is consistent with previous findings (110, 361, 389). Accelerations along the mediolateral axis provide insight into the contribution of lateral movements and may inform practitioners of postural control and stability deficits during gait tasks (73). Over time, mediolateral accelerations improved across all activities when derived from the lumbar-mounted IMUs and for jog and run activities when derived from the thoracic-mounted IMUs. Greater contributions of accelerations in the mediolateral axis have been seen post-concussion during walking tasks (389), and this persists over tasks of increasing velocity. The higher entropy scores within the concussed group are indicative of more irregular, chaotic gait, which is in accordance with previous work (300). When derived from thoracic-mounted devices, concussed athletes demonstrated greater irregularity in mediolateral accelerations for walk, jog, and run activities.

As time since concussion increased, regularity improved only during running activities and did not change when walking or jogging. Lumbar-mounted devices displayed similar findings during walking, however found no increase for jog and run activities. These scores did not change as time since concussion increased. Increased entropy scores (i.e., greater movement irregularity) and accelerations mediolaterally may highlight reduced postural control as velocity increases, resulting in more movement in the frontal plane (7). However, the clinical relevance of these changes and their association with alterations in movement proficiency, energy cost, or injury risk for team sport athletes remains unclear. Future research should aim to determine the magnitude of changes associated with reductions in movement proficiency, potentially resulting in increased injury risk or performance implications upon RTP. With regards to location differences, lumbar-mounted devices are more representative of COM and lumbo-pelvic movements (390, 391). Thoracic-mounted IMUs may provide greater insight into upper body symmetry and trunk rotation (392). Greater entropy scores seen at the thoracic region may suggest concussed athletes display more side-to-side asymmetries in the upper body whilst still maintaining relative symmetry through the lumbo-pelvic region. Collectively, the perturbations seen mediolaterally are in accordance with other authors (361, 389, 393) and highlight that sensorimotor impairments persist beyond RTP and the improvement in these metrics differs depending on the movement velocity.

Measures of vertical GRF, COM displacement, flight time, and sample entropy along the vertical axis were, in general, reduced amongst the concussed athletes. As velocity increased, greater lower leg stiffness is required to increase propulsive energy, creating a spring-like mechanism during contact with the ground (394). These findings indicate that although concussed athletes performed jogging and running activities at greater gait velocities, gait mechanics (e.g. propulsion) and regularity may be compromised. Increased gait speed is associated with a more complex, less stable gait pattern and demands greater requirements from the CNS (395). Hypoexcitability of the cortical pathways or increased intra-cortical inhibition experienced post-concussion may affect the ability of the brain to plan and coordinate movements effectively (123, 131, 188, 266, 387). This may be exacerbated as movement velocity increases due to the increasing CNS demands (395) and may result in altered muscle recruitment, leading to a less stiff musculature system (6) and a less predictable gait pattern (300). Previously concussed athletes have demonstrated quadriceps weakness and decreased peak vertical ground reaction force (PVGRF) during single-leg jump test (6). When applied to a gait specific scenario, this may then be depicted as a less 'springy' and reproducible gait (6). Reductions in vertical accelerations have been used to identify changes in movement strategies under player fatigue (171, 173, 175). Poorer neuromuscular control (e.g. while fatigued) has been shown to increase lower limb injury risk (396). Greater entropy scores have been associated with a greater relative risk of sustaining a SRC (236) and a subsequent injury following SRC (397). Similar to the mediolateral entropy outcomes presented for the thoracic and lumbar sites, increased time since concussion was associated with improvements for variables of acceleration mediolateral, acceleration vertical, and COM displacement. However, measures of vertical entropy showed little to no improvement over time for the activity of jogging for both thoracic- and lumbar-mounted IMUs. Athletes spend large portions of game time performing low intensity (e.g. jog) activities (398, 399). The between-group differences seen when performing this single-task gait assessment may contribute to the increased injury risk when athletes return to more chaotic environments (e.g. game) with added cognitive demands. These metrics of COM displacement, vertical GRF, and sample entropy for the vertical acceleration may be useful to monitor post-concussion and inform individualised rehabilitation to improve force transfer through the ground. Clinicians and practitioners should be aware of the reliability and minimal detectable change (MDC) associated with each outcome variable and device location. To determine if a change in performance is practically meaningful, practitioners should refer to the MDC of the respective measure.

The lumbar- and thoracic-mounted IMUs offered similar findings for linear measures of cadence and vertical acceleration, and the non-linear measure of sample entropy for the vertical acceleration. Differences did exist for measures of sample entropy for the mediolateral acceleration during the jog and run activities. Notably, thoracic-mounted devices show greater differences in mediolateral acceleration sample entropy when compared to lumbar-mounted devices. Due to thoracic-mounted devices being further up the kinetic chain, greater variability in upper body and trunk movements may contribute to the increased entropy scores (392). The thoracic spine may however be a better location if athletes are participating in sports that require upper-limb or trunk contributions (e.g. rugby tackle) as a lack of postural control of the trunk may lead to increased injury risk (400). Feasibility and reliability of each location should also be considered. Lumbar-mounted IMUs may be impractical for many team sport athletes as they may restrict certain sport-specific requirements (e.g. tackling). Thoracic-mounted IMUs are already utilised by many team sport athletes and pose less restrictions. However, differences in reliability and MDC requirements vary across device location and outcome measure. Practitioners should be aware of these differences prior to individual assessment. In summary, practitioners should be aware of the limitations associated with each location including their feasibility and reliability. However, our findings support the use of lumbar- or thoracic-mounted IMUs to assess gait strategies amongst concussed athletes.

The limitations of this study do need to be acknowledged. Firstly, whilst we assessed symptoms via the PCSS, the true severity of each medically diagnosed concussion was not disclosed, limiting the ability to determine if gait changes were associated with concussion severity. Secondly, this study utilised a case-control design, comparing concussed athletes with an age, sex, and weight matched control where possible. Gait strategies are unique to each individual, therefore variability between groups not attributed to concussion status may be present, particularly for the assessment of linear measures. These limitations inhibit the ability to draw causal inferences between concussion and movement strategies, thereby limiting the immediate clinical application of these findings. Future research should aim to implement within-subject, prospective designs, allowing for comparison between pre- and post-concussion gait outcomes. As we used return to run as the clearance for participation in the study, the average time of the first assessment was $5.8 \pm 5$ days. Future work should look to assess players during these light aerobic activities as soon as possible post-concussion. The clinical importance of differences is unclear and should be a focus of future research to determine the ramifications of these differences from a performance and injury perspective.

## 5.6 Conclusion

During simple walking, jogging and running tasks, concussed athletes displayed differences in gait compared to healthy controls that persisted beyond RTP, while some metrics showed no signs of recovery at 4 weeks beyond RTP. The Bayesian methods used to analyse these data allow for the true uncertainty in outcome measures to be captured across time. As gait velocity increased from walk to jog to run, concussed athletes displayed compensatory strategies as seen through linear (vertical GRF, COM displacement, flight time) and non-linear (vertical and mediolateral sample entropy) analyses. Following concussion, athletes demonstrated smaller vertical accelerations when walking and lower displacement with greater irregularity in gait.

While both lumbar- and thoracic-mounted IMUs showed differences between concussed and healthy groups, thoracic-mounted IMUs may be a more practical option for team sport athletes, given their widespread use in the field. This study illustrates some potential shortcomings of data derived from the thoracic site, but practitioners can be confident that it will still provide reliable and valid outcomes to assess potential changes in gait strategies following SRC. Findings of this study may provide clinicians and practitioners with a more comprehensive analysis of single-task gait, informing them of additional assessment requirements (e.g. dual-task gait) as well as RTP readiness. These data can be collected via thoracic-worn IMUs, during self-paced constant velocity tasks making data collection practical and feasible for team sport athletes within a training environment. Future research using within-subject design is warranted and may provide greater insight into individual recovery characteristics following SRC. Further, understanding of how these variables change when players are under fatigue would be worthwhile.

# Chapter 6

# Conclusions and Applications

## 6.1 Aims of the thesis

The aims of this thesis were to (1) Consolidate the reliability and validity of motor function assessments currently used in concussion management and further summarise their feasibility for clinicians and other end-users. (2) Assess the reliability of ankle-, lumbar-, and thoracic-mounted devices to measure gait parameters on grassed surfaces. (3) Determine if lumbar- and thoracic-mounted IMUs can detect changes in movement strategies amongst athletes returning to play following a diagnosed sports-related concussion (SRC). The studies included within this thesis aimed to highlight the limitations associated with the current clinical management of SRC and provide insight into alternative strategies that may improve current management solutions. Findings presented within this thesis provide insight into alternative solutions to monitor motor strategy changes following SRC in a manner that is objective, simple and practical for clinicians and practitioners.

Studies included within this thesis to address these aims were:

- A systematic review - Validity and reliability of methods to assess movement deficiencies following concussion
- Reliability of wearable devices for measuring gait parameters on grassed surfaces amongst healthy adults
- Movement changes persist beyond return-to-play during walking, jogging and running following sport-related concussion

How each of these studies addressed the thesis aims is outlined in the following sections.

## 6.2 Summary of studies included within this thesis

### 6.2.1 Systematic review - Validity and reliability of methods to assess movement deficiencies following concussion

Following a concussion, it appears returning athletes are at greater risk of sustaining a lower limb musculoskeletal injury or subsequent concussion (8, 99, 101, 259). Assessment of movement capabilities (e.g., walking gait, balance tasks) has highlighted lingering deficits in CNS function (73, 99, 361). Notably the implications surrounding coordination and planning of movements through the pre-motor and supplementary motor areas are believed to result in delayed communication between motor cortex and peripheral musculature (76) and have been associated with this increased injury risk (8, 99, 101, 259). Given movement issues can persist beyond clinical resolution of symptoms and may provide information on injury risk, it is important to understand the reliability and validity of motor function assessments currently used in concussion management, and further summarise their feasibility for clinicians and other end-users, to allow for informed decisions to be made. As such, a COSMIN systematic review was undertaken.

There is a multitude of instrumented and non-instrumented assessments made available to practitioners for the assessment and management of motor function following concussion. However, in most cases, clinicians utilise the Sport Concussion Assessment Tools to diagnose injury; movement changes are assessed via the non-instrumented, subjective assessments (e.g., mBESS and tandem gait) embedded within these protocols (187, 378). It is important for practitioners to understand the strengths and limitations of their current protocols as well as understand what options there are for additional assessments. Notably, the reliability, validity, and feasibility should be considered when selecting an appropriate assessment. The first study within this thesis aimed to consolidate the reliability and validity of motor assessments currently used in concussion management and further summarise their feasibility for clinicians and other-end users. In addition, this study also provided insight into the limitations associated with the current clinical management of concussion, highlighting the need for easy to administer objective assessments to better inform clinician and athletes.

#### 6.2.1.1 Key findings

1. Commonly used static balance assessments for concussion management, namely the BESS and mBESS, display insufficient (ICC <0.7) test-retest reliability and sensitivity for the detection of sports-related concussions beyond 2-weeks post-concussion. It is suspected that the simplicity of the assessments coupled with the crude nature of scoring limits their sensitivity when visible signs and symptoms dissipate, typically within 48-72 hours (70). The use of laboratory equipment such as 3D motion capture or force plates improves the test-retest reliability and sensitivity of static balance assessments. For example, the BESS reliability improved from 0.41-0.94 as a subjective assessment to 0.88-0.89 when instrumented. However, the need for expensive, specialised

equipment that require specific expertise have reduced up-take of these tools amongst clinicians and practitioners.

2. In general, static balance assessments (instrumented and non-instrumented) display poor sensitivity beyond acute stages of concussion (>2-weeks). Where possible, dynamic and gait-based assessments should be prioritised over static balance assessments. Following the most recent iteration of the SCAT6 and SCOAT6, clinicians and practitioners are to implement a timed single-task tandem gait assessment and encouraged to perform dual-task gait assessments in conjunction with static balance assessments. However, these assessments are still completed via subjective analysis, which has inherent issues with respect to inter- and intra-assessor reliability. If clinicians and practitioners choose not to implement dynamic and/or gait-based assessments, then they should be aware of the limitations associated with their selected assessment.

3. Instrumented dynamic assessments, such as a Y-Balance Test utilising lumbar mounted IMUs, demonstrate sufficient test-retest reliability and sensitivity in concussed athletes. However, this assessment displays poor specificity and may incorrectly classify healthy athletes as concussed.

4. Instrumented and non-instrumented gait assessments, in general, display sufficient test-retest reliability and sensitivity. Practitioners are encouraged to implement gait assessments where possible to assess movement capabilities post-concussion.

5. The most common gait metrics were cadence, step time, number of steps, and gait velocity. Of these, step length, step time, and velocity tended to be the most reliable with ICCs typically greater than 0.70. Cadence, velocity, accelerations in the anterior-posterior, mediolateral, and vertical axes appeared to regularly differentiate between healthy and concussed participants. As such, instrumented gait assessments should focus on these spatiotemporal gait metrics.

6. The use of non-linear analyses (e.g. Shannon entropy, sample entropy, laminarity) during IMU instrumented gait assessments show potential in identifying alterations in

movement patterns that may persist beyond clinical recovery following concussion (300, 397).

6.2.1.2 Study limitations

The methodological quality of studies investigating the reliability and/or validity of motor assessments following a sports-related concussion ranged from *inadequate* to *very good*. Most assessment types (instrumented and non-instrumented) displayed similar methodological quality scores for test-retest reliability and validity outcomes. Studies investigating test-retest reliability had the greatest differences in methodological quality for instrumented versus non-instrumented gait assessments. The quality of studies investigating the reliability of non-instrumented gait assessments ranged from *doubtful* to *adequate* compared with the *adequate* rating for instrumented assessments. It should also be noted that studies assessing the internal consistency of instrumented single-task and dual-task gait both displayed *inadequate* methodological scores. Similar trends can be seen when observing the methodological quality of studies investigating the discriminant ability of motor assessments. Instrumented and non-instrumented static assessments displayed the greatest differences with ratings of *very good* to *adequate* and *adequate,* respectively. Differences in the methodological quality may have influenced the results of individual studies included within the review. When interpreting findings and determining the best-practice assessment, researchers and clinicians should be aware of the differences in quality and refer to the methodological tables included within the review to ensure the rigor of relevant studies meets their expectations.

### 6.2.2 Reliability of wearable devices for measuring gait parameters on grassed surfaces amongst recreationally active adults

The results of the systematic review highlighted that gait assessments were the most reliable and sensitive to detecting movement deficiencies in concussed participants. It also highlighted that there were some key spatiotemporal metrics that could be used to assess movement during gait tasks. However, issues surrounding feasibility still exist. Notably, the need to perform stand-alone assessments, within controlled laboratory settings using equipment that practitioners are unlikely to have at their disposal are clear barriers. These limitations mean that whilst laboratory-based assessments provide high resolution for hypothesis testing, these assessments are unlikely to be used within practice. Athletes often wear devices embedded with IMUs which may offer clinicians and practitioners greater flexibility to collect gait data within the field. However, investigation into the reliability of linear and non-linear gait metrics derived from thoracic-mounted devices over a range of movement velocities performed on natural grassed surfaces at both set-paced and self-paced conditions is warranted.

6.2.2.1 Key findings

1. Thoracic-mounted devices displayed *moderate* to *excellent* ICCs ($r$ = 0.62 to 0.98) across all pace conditions (set- and self-paced) and movement velocities. Normalised MDCs for most variables were below 30% relative to the mean representing a detectable change. These results are similar to those seen during instrumented walking assessments performed on firm surfaces (e.g. laboratory, indoor track) ($r$ = 0.68 to 0.99) (247-249, 251).

2. Lumbar- and ankle-mounted devices displayed *moderate* to *good* ICCs for most variables. Measures of mediolateral acceleration, sample entropy, mediolateral (lumbar-mounted IMU), and step time variation had poor test-retest reliability for these locations. Normalised MDCs were below 50% of the mean for lumbar- and ankle-mounted devices.

3. This study shows that thoracic-mounted devices can reliably measure a variety of linear and non-linear gait metrics during overground gait activities performed on grass playing surfaces at varying velocities. Considering most team sport athletes already use thoracic-mounted devices to collect training and game-play data, this location may be a feasible option for practitioners if they wish to collect single-task gait data.

4. Set- and self-paced conditions displayed acceptable reliability across movement velocities. This finding provides practitioners with the ability to retrospectively analyse pre-injury gait data should they wish to.

5. Limitations associated with device anatomical location exist. Notably, ankle- and lumbar-mounted IMUs may impede on sport-specific movements (e.g. tackling, kicking

a ball). Thoracic-mounted IMUs may be the least restrictive for team sport athletes, particularly given their routine use for player load monitoring.

The results of this study highlight the potential for ankle, lumbar, and thoracic-mounted units to reliably collect spatiotemporal gait metrics over a range of velocities performed on natural grassed surfaces. Given most athletes already use thoracic-mounted IMUs, the findings of this study further support their implementation across team sports to monitor performance and/or injury rehab. Given the feasibility of this assessment, this data may provide insight into variations in gait following injury, allowing individualised rehabilitation following injury (e.g. concussion, lower-limb musculoskeletal injury). Considering the context of this assessment, determining the ability of wearable devices to differentiate between healthy and injured athletes is warranted.

### 6.2.3 Movement changes persist beyond return-to-play during walking, jogging and running following sport-related concussion

Given the issues presented in the systematic review and the results of Study 2, which highlighted that thoracic-mounted IMUs worn in tight-fitting vests can reliably capture spatiotemporal gait metrics during tasks of increasing velocities under set- and self-paced conditions. Thoracic and lumbar locations were most reliable and therefore used in this study. Lumbar-mounted IMUs have been previously used within concussion literature (115, 151, 248) and therefore acted as a point of reference for this study. The most reliable metrics were cadence, step time average, acceleration vertical, sample entropy vertical, and sample entropy mediolateral. These metrics were used in this study to determine if differences exist between concussed and healthy control participants. Notably, we sought to determine the time course of these differences and the impact of increasing velocity.

#### 6.2.3.1 Key findings

1. Compared to healthy athletes, concussed athletes demonstrated alterations in gait strategies beyond RTP timelines during instrumented single-task gait assessments of walk, jog, and run performed on natural grassed surfaces.

2. Lumbar- and thoracic-mounted IMUs both showed differences between concussed and healthy athletes for a range of linear and non-linear outcome measures.

3. Linear measures of mediolateral acceleration and vertical COM displacement displayed the largest between group differences. Specifically, concussed athletes showed a greater amount of mediolateral acceleration for both the lumbar- and thoracic-mounted IMUs, suggesting more side-to-side movements when compared with healthy controls. Whilst there is potentially a greater signal with this metric, practitioners should be aware of the poorer reliability associated with the measurement of mediolateral acceleration (thoracic: $r$ = 0.34-0.71, lumbar: $r$ = 0.52-0.58). Notably, there is an association with reduced reliability as velocity increases. As such, greater changes are needed in mediolateral accelerations for them to be deemed real.

4. Concussed athletes demonstrated greater irregularity in vertical acceleration as seen via non-linear sample entropy analyses. Between group differences in vertical sample entropy scores were exacerbated as velocity increased. It was proposed that alterations in CNS function following concussion, particularly effects on the primary motor cortex, may lead to reductions in lower-limb stiffness and affect the brain's ability to coordinate movements as velocity increases. The number of days since concussion was associated with improved entropy scores for walk and run activities, however no improvement was seen for jogging.

5. Differences between device location were most clear for the analysis of sample entropy of the mediolateral acceleration. When derived from lumbar-mounted IMUs, concussed athletes only demonstrated greater irregularity, as depicted by increased entropy scores during walk activities. However, when derived from thoracic-mounted IMUs, gait patterns were irregular for all activities. Lumbar-mounted IMUs may be more representative of lumbo-pelvic movements and COM displacement, whereas thoracic-mounted IMUs may be influenced by upper-limb and trunk movements. Therefore, greater entropy scores seen at the thoracic region may suggest less regular or predictable movements of the upper body post-concussion.

6. Lumbar- and thoracic-mounted IMUs can be useful to aid practitioners in assessing gait strategies of increasing velocities on natural grassed surfaces. When assessing

individual recovery following SRC, practitioners should be aware of the reliability associated with the IMU site and outcome variables. To determine a meaningful change in performance, practitioners should ensure changes are greater than the MDC when monitoring individual athletes.

## 6.3 Practical applications

There are several practical applications clinicians and practitioners can take away from the findings presented within this thesis.

1. Currently available assessments (e.g., BESS) offer greater ease for clinicians and practitioners compared to instrumented assessments. However, they are typically associated with poorer reliability and sensitivity outcomes beyond the acute stage. Due to this, these assessments may be useful as part of a multimodal assessment battery (e.g. SCAT6), however are not reliable as part of athlete monitoring beyond acute stages. Gait assessments seem to be the most reliable and sensitive assessments to monitor athlete recovery post-SRC, with instrumented assessments recommended over non-instrumented. Should clinicians and practitioners not have access to instruments (e.g., IMUs) yet still wish to conduct a comprehensive analysis of motor function post-concussion, the addition of a non-instrumented gait assessment as part of a multimodal assessment battery is recommended. Dual-task gait assessments may provide greater insight into underlying deficits whilst better replicating training and game demands.

2. Should clinicians and practitioners have access to wearable microtechnology (IMUs), they can be confident that ankle-, lumbar-, and thoracic-mounted IMUs can reliably extract a range of gait variables including linear measures of cadence, step time average, acceleration vertical, and vertical GRF and non-linear measures of sample entropy, vertical acceleration on natural grassed surfaces. Whilst, lumbar- and thoracic-mounted IMUs can be used to detect changes in gait strategies following concussion over velocities up to 6 $m \cdot s^{-1}$.

3. When applied in the team sport environment, clinicians and practitioners can easily collect data from thoracic- or lumbar-mounted IMUs over any constant paced activity up to ~6 $m \cdot s^{-1}$ performed during training. Most team sport athletes already use thoracic-mounted IMUs to monitor training and game loads. Therefore, these devices are readily available to help assess gait strategies following SRC. The simplicity of the assessment (single-task gait) allows practitioners and/or coaching staff to assess gait strategies during activities athletes already participate in. For example, where players are normally instrumented in training, practitioners can collect gait data from sections of the warm-up where athletes perform constant velocity tasks over 20-30 m. This allows objective data to be collected without burdening to players or practitioners.

4. Considering athletes perform instrumented gait activities throughout training, data can also be collected retrospectively (e.g., prior to concussion), allowing pre- and post-concussion gait analysis. This would enable practitioners to make more informed decisions regarding individual recovery timelines and RTP readiness.

5. When derived from thoracic-mounted IMUs, linear measures of mediolateral acceleration and vertical COM displacement are most sensitive to changes in gait strategies post-concussion. The addition of non-linear analyses may provide greater insight into the regularity of movements post-concussion, particularly vertical accelerations. However, practitioners and clinicians should be aware of the limitations associated with non-linear analysis. Sample entropy measures the regularity of time-series data and is therefore reliant on having enough data points of consistent velocity (341). Within the context of gait, to accurately assess regularity, a large quantity of data points are required. This is typically seen through long-duration gait assessments where multiple strides are recorded, resulting in larger data sets. Shorter data sets can be problematic as there may be an insufficient number of data points to accurately assess the regularity of movements occurring (293, 341). As participants increase their velocity, fewer data points are collected, reducing the amount of information provided to determine regularity of gait cycles. Regular training activities like aerobic-based rehab (e.g., graded running tasks) or steady-state activities during the warm-up, would provide a good setting to capture (potentially) thousands of gait cycles to quantify sample entropy.

6. The application of these findings may extend beyond concussion management. Given the incidence of lower-limb injuries amongst team sport athletes and the changes in function (e.g., vertical stiffness) that may occur post-injury, they may be useful in monitoring injury rehabilitation. Specifically, clinicians and practitioners may benefit from the additional analysis of pre- and post-injury gait data derived from ankle-, lumbar-, or thoracic-mounted devices to better inform recovery and RTP timelines for a variety of lower-limb injuries. Further, certain metrics (e.g., vertical stiffness, ground reaction forces) may be useful in regular performance monitoring without the need for standalone assessments.

7. Although the extraction of the gait parameters in this study required accessing and analysing the raw data, practitioners can obtain vertical and mediolateral acceleration profiles directly from the proprietary software of the microtechnology providers, which can be easily monitored during and beyond RTP. These metrics may serve as a first check to see whether more in-depth analysis is required to understand more granular metrics like sample entropy, centre of mass displacement, and cadence.

## 6.4 Actionable guidelines

Highlighted in this section are actionable guidelines clinicians and practitioners may use to assist with the selection and implementation of gait assessments investigated within this thesis.

### 6.4.1 Testing times and baseline assessments

1. Within-subject baseline data collection should be implemented where feasible for clinicians. Due to the simplicity of the assessment, data can be collected during any straight-line continuous walk, jog, or run task at a constant, self-paced velocity.

2. Should a clinician be unable to obtain baseline data prior to a concussion occurring, they may be able to obtain appropriate data retrospectively if athletes are commonly fitted with thoracic or lumbar mounted IMUs. If this occurs, clinicians should be aware of the activity (walk, jog, run) and speed used for the baseline data and aim to replicate those parameters when conducting subsequent assessments.

3. Should within-subject comparison be unachievable, clinicians are advised to match concussed players with an age, sex, and weight matched control where possible.

4. Post-concussion testing should be conducted as soon as possible once the athlete has been cleared for RTR activities.

5. Testing times should follow the same time frames as implemented in the studies. Clinicians should be reminded that the discriminant ability of thoracic and/or lumbar mounted IMUs has not been assessed beyond four-weeks post RTP. Should clinicians decide to assess athletes beyond this time point, caution must be applied when interpreting results and should not be used in isolation to determine an athlete's match readiness or recovery.

### 6.4.2 Protocol

1. Any unobstructed continuous straight-line walk, jog, or run task over a 20-meter distance performed on a grassed surface wearing the same footwear (e.g., boots) can be used for pre- and post-concussion comparisons.

2. Clinicians should prioritise the assessment of jog and run activities where possible. Higher velocity activities such as jogging and running, displayed greater differences compared to controls for both thoracic and lumbar mounted IMUs. In addition, these activities better replicated the demands of the game.

3. The protocol used in this thesis was designed to replicate distances commonly performed during training such as during the warm-up. Should clinicians wish to regularly collect data, incorporating walk, jog, and run tasks within the warmup may be a suitable option that does not impede training activities.

### 6.4.3 Outcome measures

1. Linear measures such as cadence, step length, and vertical accelerations may not be sensitive enough to detect subclinical changes in movement strategies following

concussion in isolation. The gross nature of these outcome measures likely limits their ability to detect nuanced changes in movement strategies which may contribute to the increased injury risk seen in this population.

2. Linear measures of mediolateral accelerations derived from thoracic and/or lumbar mounted IMUs are a better option for clinicians to detect changes in movement strategies and should be combined with measures of cadence and vertical accelerations to allow a comprehensive assessment of gait. Both sites displayed increased mediolateral accelerations amongst concussed athletes across all activities. Most activities saw an improvement in these outcome measures as time since concussion increased.

3. Most wearable technology software has linear metrics, such as accelerometer load measures in individual planes, readily available for users. This option provides clinicians with a quick and easy to digest interpretation of each athlete's movement strategies without requiring significant time commitments to analyse and interpret data.

4. Non-linear metrics of vertical and mediolateral sample entropy were effective at differentiating between concussed and non-concussed athletes for all activities and device locations. Increased entropy scores may be indicative of a less coordinated system as seen through increased irregularity of gait. This may therefore provide a more comprehensive insight into movement strategies following concussion.

5. Where the resources and availability permits, clinicians should aim to analyse non-linear measures of sample entropy in conjunction with linear measures of gait.

### 6.4.4 Variations in IMU site

1. Clinicians should be aware of the inherent reliability, validity, and feasibility associated with each device location and select the best option that suits their needs and capabilities.

2. Lumbar mounted IMUs tend to better replicate COM movements. This is particularly evident when looking at linear and non-linear mediolateral accelerations. Clinicians should therefore be aware of this when interpreting findings. This device location also poses limitations with regards to feasibility, particularly for team sport athletes involved in contact events (e.g., rugby tackle). Should this location be selected, caution is advised if wearing it whilst performing sport-specific activities.

3. Whilst thoracic-mounted IMUs may circumvent some of the limitations associated with the feasibility of lumbar-mounted devices, their distance up the kinetic chain may attenuate the sensitivity of some metrics when deriving forces from ground contact. Thoracic mounted IMUs tend to better replicate upper trunk movements and the sites typically affected during concussive injuries (e.g., head and neck). Clinicians should be aware of these factors when interpreting findings.

## 6.5 Conclusion and future research

In summary, the findings presented in this thesis concluded that 1) Whilst feasible, currently used motor assessments for concussion management may lack the required sensitivity to detect potential motor deficits beyond acute (7-days) stages. 2) Ankle-, lumbar-, and thoracic-mounted IMUs can reliably collect linear and non-linear measures of gait during walk, jog, and run activities performed on natural grassed surfaces. 3) Lumbar- and thoracic-mounted IMUs show changes in gait strategies that persist beyond RTP clearance, with some showing no improvement up to ~ 2 months post-concussion.

The availability, portability, and low-cost associated with IMUs, in conjunction with the findings presented in this thesis, highlight the potential advantages these devices have over isolated traditional clinical assessments to objectively monitor motor strategy changes post-concussion. Most professional team sport athletes already use thoracic-mounted IMUs to monitor training and game loads (175). Considering these devices are readily available, data can be collected within a training environment without disrupting players or staff. Having the ability to objectively monitor gait strategies post-concussion allows for a more rounded assessment of motor function, increasing the likelihood of detecting any underlying abnormalities. Movement deficiencies are linked to increased injury risk post-concussion

(396), which may be improved via specific neuromuscular training (401, 402) during and following graduated RTP. As such, practitioners are encouraged to use IMU devices to monitor the movement characteristics following SRC to support decision making and inform rehabilitation strategies.

The studies within this thesis concluded that concussed athletes may demonstrate variations in gait strategies upon RTP when performing single-task walk, jog, and run activities. However, team sport environments are chaotic and require varying motor and cognitive demands simultaneously. The cost of these demands may also fluctuate over the duration of a game as fatigue increases. Future research should therefore aim to investigate: 1) Within-subject gait pre- and post-concussion for males and females, allowing more individualised analysis of motor function post-concussion and a more well-rounded representation of potential sex-related differences. 2) Assessment of gait post-concussion at speeds above 6 $m \cdot s^{-1}$. Evidence of alterations in gait strategies at higher speeds may highlight modifications linked to changes in performance. 3) The inclusion of dual-task assessments. Evidence suggests greater abnormalities in movement strategies may be associated with dual-task events. This thesis was limited in its assessment of single-task gait. Including both single- and dual-task constraints will provide a more wholistic overview of alterations in gait strategies following concussion. 4) The effect of acute or residual fatigue on gait strategies pre- and post-concussion. Notably, it is important to understand whether athletes display greater compensatory strategies when fatigued following return from concussion than they did pre-concussion.

# References


1. Dessy AM, Yuk FJ, Maniya AY, Gometz A, Rasouli JJ, Lovell MR, et al. Review of Assessment Scales for Diagnosing and Monitoring Sports-related Concussion. Cureus. 2017.
2. Ferry B, DeCastro A. Concussion. StatPearls [Internet]. 2019.
3. Martini DN, Broglio SP. Long-term effects of sport concussion on cognitive and motor performance: a review. International journal of psychophysiology. 2018;132:25–30.
4. Cross M, Kemp S, Smith A, Trewartha G, Stokes K. Professional Rugby Union players have a 60% greater risk of time loss injury after concussion: a 2-season prospective study of clinical outcomes. British journal of sports medicine. 2016;50(15):926–31.
5. Buckley TA. Concussion research: Moving beyond the natural history. Journal of sport and health science. 2021;10(2):111.
6. DuBose DF, Herman DC, Jones DL, Tillman SM, Clugston JR, Pass A, et al. Lower extremity stiffness changes following concussion in collegiate football players. Medicine and science in sports and exercise. 2017;49(1):167.
7. Howell DR, Kirkwood MW, Provance A, Iverson GL, Meehan III WP. Using concurrent gait and cognitive assessments to identify impairments after concussion: a narrative review. Concussion. 2018;3(1):CNC54.
8. Howell DR, Buckley TA, Lynall RC, Meehan III WP. Worsening dual-task gait costs after concussion and their association with subsequent sport-related injury. Journal of neurotrauma. 2018;35(14):1630–6.
9. Smulligan KL, Carry P, Wingerson MJ, Donahue C, Wilson JC, Howell DR. Clinical Measures Associated with Subsequent Injury after Adolescent Concussion: A Prospective Cohort Study. Medicine & Science in Sports & Exercise. 2023:10.1249.
10. Langer L, Levy C, Bayley M. Increasing incidence of concussion: true epidemic or better recognition? The Journal of head trauma rehabilitation. 2020;35(1):E60–E6.
11. Fino PC, Becker LN, Fino NF, Griesemer B, Goforth M, Brolinson PG. Effects of recent concussion and injury history on instantaneous relative risk of lower extremity injury in Division I collegiate athletes. Clinical journal of sport medicine. 2019;29(3):218–23.
12. Johnston W, Coughlan GF, Caulfield B. Challenging concussed athletes: the future of balance assessment in concussion. QJM: An International Journal of Medicine. 2017;110(12):779–83.
13. Langlois JA, Rutland-Brown W, Wald MM. The epidemiology and impact of traumatic brain injury: a brief overview. The Journal of head trauma rehabilitation. 2006;21(5):375–8.
14. Veliz P, Eckner JT, Zdroik J, Schulenberg JE. Lifetime prevalence of self-reported concussion among adolescents involved in competitive sports: a national US study. Journal of Adolescent Health. 2019;64(2):272–5.
15. Antrobus MR, Brazier J, Stebbings GK, Day SH, Heffernan SM, Kilduff LP, et al. Genetic factors that could affect concussion risk in elite rugby. Sports. 2021;9(2):19.
16. Australian Government ASC. AusPlay 2020 [Available from: https://www.clearinghouseforsport.gov.au/kb/sport-participation-in-australia/participation-statistics-and-trends.
17. Pfister T, Pfister K, Hagel B, Ghali WA, Ronksley PE. The incidence of concussion in youth sports: a systematic review and meta-analysis. British journal of sports medicine. 2016;50(5):292–7.
18. Kung SM, Suksreephaisan TK, Perry BG, Palmer BR, Page RA. The effects of anticipation and visual and sensory performance on concussion risk in sport: a review. Sports Medicine-Open. 2020;6(1):1–14.
19. West SW, Starling L, Kemp S, Williams S, Cross M, Taylor A, et al. Trends in match injury risk in professional male rugby union: a 16-season review of 10 851 match injuries in the English Premiership (2002–2019): the Professional Rugby Injury Surveillance Project. British journal of sports medicine. 2020.

20. West SW, Cross M, Trewartha G, Taylor A, Brooks J, Kemp S, et al. Trends in match concussion incidence and return-to-play time in male professional Rugby Union: A 16-season prospective cohort study. Brain injury. 2021;35(10):1235–44.
21. Lincoln AE, Caswell SV, Almquist JL, Dunn RE, Norris JB, Hinton RY. Trends in concussion incidence in high school sports: a prospective 11-year study. The American journal of sports medicine. 2011;39(5):958–63.
22. Longworth T, McDonald A, Cunningham C, Khan H, Fitzpatrick J. Do rugby league players under-report concussion symptoms? A cross-sectional study of elite teams based in Australia. BMJ Open Sport & Exercise Medicine. 2021;7(1):e000860.
23. Eastwood D, Owen C, Phillips G, Williams S, Brown J, Gardner AJ, et al. Incidence of concussion in men's Super League, Championship, and Academy rugby league matches between 2016 and 2022. Journal of science and medicine in sport. 2023;26(10):539–44.
24. McLeod S, Tucker R, Edwards S, Jones B, Page G, Spiegelhalter M, et al. A case-control study of tackle based head impact event (HIE) risk factors from the first three seasons of the National Rugby League Women's competition. Frontiers in Sports and Active Living. 2023;5:1080356.
25. Gardner A, Iverson GL, Levi CR, Schofield PW, Kay-Lambkin F, Kohler RMN, et al. A systematic review of concussion in rugby league. British Journal of Sports Medicine. 2015;49(8):495–8.
26. King DA, Hume PA, Milburn PD, Guttenbeil D. Match and training injuries in rugby league: a review of published studies. Sports Medicine. 2010;40:163–78.
27. Williams S, Robertson C, Starling L, McKay C, West S, Brown J, et al. Injuries in Elite Men's Rugby Union: An Updated (2012–2020) Meta-Analysis of 11,620 Match and Training Injuries. Sports Medicine. 2022;52(5):1127–40.
28. Dave U, Kinderknecht J, Cheng J, Santiago K, Jivanelli B, Ling DI. Systematic review and meta-analysis of sex-based differences for concussion incidence in soccer. The Physician and sportsmedicine. 2021:1–9.
29. Marar M, McIlvain NM, Fields SK, Comstock RD. Epidemiology of Concussions Among United States High School Athletes in 20 Sports. The American Journal of Sports Medicine. 2012;40(4):747–55.
30. Spiegelhalter M, Scantlebury S, Heyward O, Hendricks S, Cummins C, Gardner AJ, et al. The propensity of non-concussive and concussive head contacts during elite-level women's rugby league matches: A prospective analysis of over 14,000 tackle events. Journal of science and medicine in sport. 2023;26(3):195–201.
31. Gardner AJ, Iverson GL, Quinn TN, Makdissi M, Levi CR, Shultz SR, et al. A preliminary video analysis of concussion in the National Rugby League. Brain injury. 2015;29(10):1182–5.
32. Gardner AJ, Howell DR, Iverson GL. A video review of multiple concussion signs in National Rugby League match play. Sports Medicine - Open. 2018;4(1):1–N.PAG.
33. King D, Hume PA, Clark T. Nature of Tackles That Result in Injury in Professional Rugby League. Research in Sports Medicine. 2012;20(2):86–104.
34. Dai JB, Li AY, Haider SF, Tomaselli R, Gometz A, Sobotka S, et al. Effects of game characteristics and player positions on concussion incidence and severity in professional football. Orthopaedic journal of sports medicine. 2018;6(12):2325967118815448.
35. Mc Fie S, Brown J, Hendricks S, Posthumus M, Readhead C, Lambert M, et al. Incidence and factors associated with concussion injuries at the 2011 to 2014 South African Rugby Union Youth Week Tournaments. Clinical journal of sport medicine. 2016;26(5):398–404.
36. Chéradame J, Piscione J, Carling C, Guinoiseau J-P, Dufour B, Jacqmin-Gadda H, et al. Incidence and Risk Factors in Concussion Events: A 5-Season Study in the French Top 14 Rugby Union Championship. The American Journal of Sports Medicine. 2021;49(7):1921–8.
37. Batista-García-Ramó K, Fernández-Verdecia CI. What we know about the brain structure–function relationship. Behavioral Sciences. 2018;8(4):39.
38. Khan GM. The Biology of Brain: An Insight into the Human Brain. Springer International Publishing; 2018. p. 9–28.

39. Alisafaei F, Gong Z, Johnson VE, Dollé J-P, Smith DH, Shenoy VB. Mechanisms of local stress amplification in axons near the gray-white matter interface. Biophysical Journal. 2020;119(7):1290–300.
40. Tierney G. Concussion biomechanics, head acceleration exposure and brain injury criteria in sport: a review. Sports biomechanics. 2022:1–29.
41. Meaney DF, Smith DH. Biomechanics of concussion. Clinics in sports medicine. 2011;30(1):19–31.
42. Bilston LE. Brain tissue mechanical properties. Biomechanics of the Brain. 2019:71–95.
43. Mercadante AA, Tadi P. Neuroanatomy, gray matter. 2020.
44. Budday S, Nay R, De Rooij R, Steinmann P, Wyrobek T, Ovaert TC, et al. Mechanical properties of gray and white matter brain tissue by indentation. Journal of the mechanical behavior of biomedical materials. 2015;46:318–30.
45. Shaw NA. The neurophysiology of concussion. Progress in neurobiology. 2002;67(4):281–344.
46. Barth JT, Freeman JR, Broshek DK, Varney RN. Acceleration-deceleration sport-related concussion: the gravity of it all. Journal of athletic training. 2001;36(3):253.
47. Ellis MJ, Leddy J, Willer B. Multi-disciplinary management of athletes with post-concussion syndrome: an evolving pathophysiological approach. Frontiers in neurology. 2016;7:136.
48. Howell DR, Southard J. The Molecular Pathophysiology of Concussion. Clinics in Sports Medicine. 2021;40(1):39–51.
49. Duckworth JL, Grimes J, Ling GS. Pathophysiology of battlefield associated traumatic brain injury. Pathophysiology. 2013;20(1):23–30.
50. Gurdjian ES. Prevention and Mitigation of Injuries: Chapter 2. Neurosurgery. 1972;19:43–57.
51. Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. Exercise and sport sciences reviews. 2011;39(1):4–11.
52. Ono K, Kanno M. Influences of the physical parameters on the risk to neck injuries in low impact speed rear-end collisions. Accident Analysis & Prevention. 1996;28(4):493–9.
53. Pellman EJ, Viano DC, Tucker AM, Casson IR. Concussion in professional football: Location and direction of helmet impacts—Part 2. Neurosurgery. 2003;53(6):1328–41.
54. Post A, Hoshizaki TB, Gilchrist MD, Cusimano MD. Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. Journal of biomechanics. 2017;61:183–92.
55. Zhang L, Yang KH, King AI. A Proposed Injury Threshold for Mild Traumatic Brain Injury. Journal of Biomechanical Engineering. 2004;126(2):226–36.
56. Post A, Blaine Hoshizaki T. Rotational acceleration, brain tissue strain, and the relationship to concussion. Journal of biomechanical engineering. 2015;137(3):030801.
57. Giza CC, Hovda DA. The new neurometabolic cascade of concussion. Neurosurgery. 2014;75(0 4):S24.
58. Danton GH, Dietrich WD. Neuroprotection: Where Are We Going? From Neuroscience To Neurology: Elsevier; 2005. p. 237–65.
59. Ferry B, DeCastro A. Concussion. 2019.
60. Jin X-T, Smith Y. Localization and functions of kainate receptors in the basal ganglia. Kainate Receptors: Novel Signaling Insights. 2011:27–37.
61. Royo M, Escolano BA, Madrigal MP, Jurado S. AMPA receptor function in hypothalamic synapses. Frontiers in Synaptic Neuroscience. 2022;14:833449.
62. Jewett BE, Thapa B. Physiology, NMDA receptor. StatPearls [Internet]: StatPearls Publishing; 2022.
63. Johnson VE, Weber MT, Xiao R, Cullen DK, Meaney DF, Stewart W, et al. Mechanical disruption of the blood–brain barrier following experimental concussion. Acta neuropathologica. 2018;135:711–26.

64. Valovich McLeod TC, Hale TD. Vestibular and balance issues following sport-related concussion. Brain injury. 2015;29(2):175–84.
65. Bonke EM, Southard J, Buckley TA, Reinsberger C, Koerte IK, Howell DR. The effects of repetitive head impacts on postural control: A systematic review. Journal of Science and Medicine in Sport. 2020.
66. Mullally WJ. Concussion. The American journal of medicine. 2017;130(8):885–92.
67. Chancellor SE, Franz ES, Minaeva OV, Goldstein LE, editors. Pathophysiology of concussion. Seminars in pediatric neurology; 2019: Elsevier.
68. Kelly JP. Loss of consciousness: pathophysiology and implications in grading and safe return to play. Journal of Athletic Training. 2001;36(3):249.
69. Barlow KM. Postconcussion syndrome: a review. Journal of child neurology. 2016;31(1):57–67.
70. Murray N, Salvatore A, Powell D, Reed-Jones R. Reliability and validity evidence of multiple balance assessments in athletes with a concussion. Journal of athletic training. 2014;49(4):540–9.
71. Scott OF, Bubna M, Boyko E, Hunt C, Kristman VL, Gargaro J, et al. Characterizing the profiles of patients with acute concussion versus prolonged post-concussion symptoms in Ontario. Scientific reports. 2023;13(1):17955.
72. Ho RA, Hall GB, Noseworthy MD, Dematteo C. Post-concussive depression: evaluating depressive symptoms following concussion in adolescents and its effects on executive function. Brain Injury. 2020;34(4):520–7.
73. Buckley TA, Oldham JR, Caccese JB. Postural control deficits identify lingering post-concussion neurological deficits. Journal of sport and health science. 2016;5(1):61–9.
74. Howell DR, O'Brien MJ, Raghuram A, Shah AS, Meehan III WP. Near point of convergence and gait deficits in adolescents after sport-related concussion. Clinical journal of sport medicine. 2018;28(3):262–7.
75. Oldham JR, Howell DR, Knight CA, Crenshaw JR, Buckley TA. Gait Performance Is Associated with Subsequent Lower Extremity Injury following Concussion. Medicine and science in sports and exercise. 2020;52(11):2279–85.
76. Wilkerson GB, Grooms DR, Acocello SN. Neuromechanical considerations for postconcussion musculoskeletal injury risk management. Current sports medicine reports. 2017;16(6):419–27.
77. Ludwig R, D'Silva L, Vaduvathiriyan P, Rippee MA, Siengsukon C. Sleep disturbances in the acute stage of concussion are associated with poorer long-term recovery: A systematic review. Pm&r. 2020;12(5):500–11.
78. Mollica A, Safavifar F, Fralick M, Giacobbe P, Lipsman N, Burke MJ. Transcranial Magnetic Stimulation for the Treatment of Concussion: A Systematic Review. Neuromodulation: Technology at the Neural Interface. 2020.
79. Iverson GL. Network analysis and precision rehabilitation for the post-concussion syndrome. Frontiers in neurology. 2019;10:489.
80. Biagianti B, Stocchetti N, Brambilla P, Van Vleet T. Brain dysfunction underlying prolonged post-concussive syndrome: A systematic review. Journal of affective disorders. 2020;262:71–6.
81. McLendon LA, Kralik SF, Grayson PA, Golomb MR. The controversial second impact syndrome: a review of the literature. Pediatric neurology. 2016;62:9–17.
82. Commission AS. Concussion in Australian Sport 2024 [Available from: https://www.concussioninsport.gov.au/.
83. Tierney GJ, Kuo C, Wu L, Weaving D, Camarillo D. Analysis of head acceleration events in collegiate-level American football: a combination of qualitative video analysis and in-vivo head kinematic measurement. Journal of Biomechanics. 2020;110:109969.
84. Tooby J, Weaving D, Al-Dawoud M, Tierney G. Quantification of head acceleration events in rugby league: an instrumented mouthguard and video analysis pilot study. Sensors. 2022;22(2):584.
85. McAllister T, McCrea M. Long-term cognitive and neuropsychiatric consequences of repetitive concussion and head-impact exposure. Journal of athletic training. 2017;52(3):309–17.

86. Black SE, Follmer B, Mezzarane RA, Pearcey GEP, Sun Y, Zehr EP. Exposure to impacts across a competitive rugby season impairs balance and neuromuscular function in female rugby athletes. Bmj Open Sport & Exercise Medicine. 2020;6(1).
87. Kuzminski SJ, Clark MD, Fraser MA, Haswell CC, Morey RA, Liu C, et al. White Matter Changes Related to Subconcussive Impact Frequency during a Single Season of High School Football. American Journal of Neuroradiology. 2018;39(2):245–51.
88. McAllister TW, Flashman LA, Maerlender A, Greenwald RM, Beckwith JG, Tosteson TD, et al. Cognitive effects of one season of head impacts in a cohort of collegiate contact sport athletes. Neurology. 2012;78(22):1777–84.
89. McAllister TW, Ford JC, Flashman LA, Maerlender A, Greenwald RM, Beckwith JG, et al. Effect of head impacts on diffusivity measures in a cohort of collegiate contact sport athletes. Neurology. 2014;82(1):63–9.
90. TalavageThomas M, NaumanEric A, BreedloveEvan L, DyeAnne E, MorigakiKatherine E, LeverenzLarry J. Functionally-detected cognitive impairment in high school football players without clinically-diagnosed concussion. Journal of neurotrauma. 2014.
91. Horak FB. Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? Age and ageing. 2006;35(suppl_2):ii7–ii11.
92. Daneshvar DH, Nair ES, Baucom ZH, Rasch A, Abdolmohammadi B, Uretsky M, et al. Leveraging football accelerometer data to quantify associations between repetitive head impacts and chronic traumatic encephalopathy in males. Nature communications. 2023;14(1):3470.
93. Tadmor D, Till K, Phillips G, Brown J, Fairbank L, Hendricks S, et al. I won't let you down; why 20% of Men's and Women's Super League players underreported suspected concussions. Journal of science and medicine in sport. 2023;26(12):688–93.
94. Cobb BR, Tyson AM, Rowson S. Head acceleration measurement techniques: Reliability of angular rate sensor data in helmeted impact testing. Proceedings of the Institution of Mechanical Engineers Part P-Journal of Sports Engineering and Technology. 2018;232(2):176–81.
95. Cobb BR, Urban JE, Davenport EM, Rowson S, Duma SM, Maldjian JA, et al. Head impact exposure in youth football: elementary school ages 9–12 years and the effect of practice structure. Annals of biomedical engineering. 2013;41:2463–73.
96. Jones B, Tooby J, Weaving D, Till K, Owen C, Begonia M, et al. Ready for impact? A validity and feasibility study of instrumented mouthguards (iMGs). British journal of sports medicine. 2022;56(20):1171–9.
97. Hendricks S. Tackle injury epidemiology and performance in rugb. 2021.
98. Tooby J, Woodward J, Tucker R, Jones B, Falvey É, Salmon D, et al. Instrumented mouthguards in elite-level men’s and women’s rugby union: the incidence and propensity of head acceleration events in matches. Sports medicine. 2024;54(5):1327–38.
99. Howell DR, Lynall RC, Buckley TA, Herman DC. Neuromuscular control deficits and the risk of subsequent injury after a concussion: a scoping review. Sports medicine. 2018;48(5):1097–115.
100. Brooks MA, Peterson K, Biese K, Sanfilippo J, Heiderscheit BC, Bell DR. Concussion increases odds of sustaining a lower extremity musculoskeletal injury after return to play among collegiate athletes. The American journal of sports medicine. 2016;44(3):742–7.
101. Herman DC, Jones D, Harrison A, Moser M, Tillman S, Farmer K, et al. Concussion may increase the risk of subsequent lower extremity musculoskeletal injury in collegiate athletes. Sports medicine. 2017;47(5):1003–10.
102. Reneker JC, Babl R, Flowers MM. History of concussion and risk of subsequent injury in athletes and service members: a systematic review and meta-analysis. Musculoskeletal science and practice. 2019;42:173–85.
103. Garcia G-GP, Czerniak LL, Lavieri MS, Liebel SW, Van Pelt KL, Pasquina PF, et al. Estimating the relationship between the symptom-free waiting period and injury rates after return-to-play from concussion: a simulation analysis using CARE Consortium data. Sports medicine. 2023:1–16.

104. Makdissi M, McCrory P, Ugoni A, Darby D, Brukner P. A prospective study of postconcussive outcomes after return to play in Australian football. The American journal of sports medicine. 2009;37(5):877–83.
105. Nyberg G, Mossberg KH, Tegner Y, Lysholm J. Subsequent traumatic injuries after a concussion in elite ice hockey: a study over 28 years. Current Research: Concussion. 2015;2(3):109–12.
106. Taylor KM, Kioumourtzoglou M-A, Clover J, Coull BA, Dennerlein JT, Bellinger DC, et al. Concussion history and cognitive function in a large cohort of adolescent athletes. The American journal of sports medicine. 2018;46(13):3262–70.
107. Kleiner M, Wong L, Dubé A, Wnuk K, Hunter SW, Graham LJ. Dual-Task Assessment Protocols in Concussion Assessment: A Systematic Literature Review. Journal of Orthopaedic & Sports Physical Therapy. 2018;48(2):87–103.
108. Lee H, Sullivan SJ, Schneiders AG. The use of the dual-task paradigm in detecting gait performance deficits following a sports-related concussion: a systematic review and meta-analysis. Journal of science and Medicine in Sport. 2013;16(1):2–7.
109. Pearson KG. Generating the walking gait: role of sensory feedback. Progress in brain research. 2004;143:123–9.
110. Howell DR, Stracciolini A, Geminiani E, Meehan III WP. Dual-task gait differences in female and male adolescents following sport-related concussion. Gait & posture. 2017;54:284–9.
111. Wilkerson GB, Nabhan DC, Prusmack CJ, Moreau WJ. Detection of Persisting Concussion Effects on Neuromechanical Responsiveness. Medicine and science in sports and exercise. 2018;50(9):1750–6.
112. McPherson AL, Nagai T, Webster KE, Hewett TE. Musculoskeletal injury risk after sport-related concussion: a systematic review and meta-analysis. The American journal of sports medicine. 2019;47(7):1754–62.
113. Howell DR, Osternig LR, Chou LS. Consistency and cost of dual-task gait balance measure in healthy adolescents and young adults. Gait & Posture. 2016;49:176–80.
114. Van Deventer KA, Seehusen CN, Walker GA, Wilson JC, Howell DR. The diagnostic and prognostic utility of the dual-task tandem gait test for pediatric concussion. Journal of Sport and Health Science. 2021;10(2):131–7.
115. Howell DR, Buckley TA, Berkstresser B, Wang F, Meehan WP. Identification of Postconcussion Dual-Task Gait Abnormalities Using Normative Reference Values. Journal of Applied Biomechanics. 2019;35(4):290–6.
116. Fait P, Swaine B, Cantin J-F, Leblond J, McFadyen BJ. Altered integrated locomotor and cognitive function in elite athletes 30 days postconcussion: a preliminary study. The Journal of head trauma rehabilitation. 2013;28(4):293–301.
117. Leddy J, Baker JG, Haider MN, Hinds A, Willer B. A physiological approach to prolonged recovery from sport-related concussion. Journal of athletic training. 2017;52(3):299–308.
118. Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. Med Sci Sports Exerc. 1992;24(1):108–15.
119. Yip DW, Lui F. Physiology, motor cortical. 2019.
120. Sears SM, Hewett SJ. Influence of glutamate and GABA transport on brain excitatory/inhibitory balance. Experimental Biology and Medicine. 2021;246(9):1069–83.
121. Miller NR, Yasen AL, Maynard LF, Chou L-S, Howell DR, Christie AD. Acute and longitudinal changes in motor cortex function following mild traumatic brain injury. Brain injury. 2014;28(10):1270–6.
122. Tremblay S, De Beaumont L, Lassonde M, Théoret H. Evidence for the specificity of intracortical inhibitory dysfunction in asymptomatic concussed athletes. Journal of Neurotrauma. 2011;28(4):493–502.
123. Powers KC, Cinelli ME, Kalmar JM. Cortical hypoexcitability persists beyond the symptomatic phase of a concussion. Brain injury. 2014;28(4):465–71.

124. Yasen AL, Lim MM, Weymann KB, Christie AD. Excitability, Inhibition, and Neurotransmitter Levels in the Motor Cortex of Symptomatic and Asymptomatic Individuals Following Mild Traumatic Brain Injury. Frontiers in Neurology. 2020;11.
125. Livingston SC, Saliba EN, Goodkin HP, Barth JT, Hertel JN, Ingersoll CD. A preliminary investigation of motor evoked potential abnormalities following sport-related concussion. Brain injury. 2010;24(6):904–13.
126. Chistyakov A, Soustiel J, Hafner H, Trubnik M, Levy G, Feinsod M. Excitatory and inhibitory corticospinal responses to transcranial magnetic stimulation in patients with minor to moderate head injury. Journal of Neurology, Neurosurgery & Psychiatry. 2001;70(5):580–7.
127. De Beaumont L, Theoret H, Mongeon D, Messier J, Leclerc S, Tremblay S, et al. Brain function decline in healthy retired athletes who sustained their last sports concussion in early adulthood. Brain. 2009;132(3):695–708.
128. Reed N, Taha T, Keightley ML. Concussion and strength performance in youth ice hockey players. British Journal of Sports Medicine. 2013;47(5):e1–e.
129. Edwards EK, Christie AD. Assessment of motor cortex excitability and inhibition during a cognitive task in individuals with concussion. Brain injury. 2017;31(10):1348–55.
130. De Beaumont L, Lassonde M, Leclerc S, Théoret H. Long-term and cumulative effects of sports concussion on motor cortex inhibition. Neurosurgery. 2007;61(2):329–37.
131. De Beaumont L, Mongeon D, Tremblay S, Messier J, Prince F, Leclerc S, et al. Persistent motor system abnormalities in formerly concussed athletes. Journal of athletic training. 2011;46(3):234–40.
132. Watanabe M, Maemura K, Kanbara K, Tamayama T, Hayasaki H. GABA and GABA receptors in the central nervous system and other organs. International review of cytology. 2002;213:1–47.
133. Patricios JS, Schneider KJ, Dvorak J, Ahmed OH, Blauwet C, Cantu RC, et al. Consensus statement on concussion in sport: the 6th International Conference on Concussion in Sport–Amsterdam, October 2022. British Journal of Sports Medicine. 2023;57(11):695–711.
134. Echemendia RJ, Brett BL, Broglio S, Davis GA, Giza CC, Guskiewicz KM, et al. Introducing the Sport Concussion Assessment Tool 6 (SCAT6). British journal of sports medicine. 2023;57(11):619–21.
135. Rodrigues MR, Slimovitch M, Chilingaryan G, Levin MF. Does the Finger-to-Nose Test measure upper limb coordination in chronic stroke? Journal of neuroengineering and rehabilitation. 2017;14(1):1–11.
136. Brown JA, Dalecki M, Hughes C, Macpherson AK, Sergio LE. Cognitive-motor integration deficits in young adult athletes following concussion. BMC sports science, medicine and rehabilitation. 2015;7(1):1–12.
137. Alsalaheen B, McClafferty A, Haines J, Smith L, Yorke A. Reference values for the balance error scoring system in adolescents. Brain Injury. 2016;30(7):914–8.
138. Wilkins JC, McLeod TCV, Perrin DH, Gansneder BM. Performance on the balance error scoring system decreases after fatigue. Journal of athletic training. 2004;39(2):156.
139. Fino PC, Parrington L, Pitt W, Martini DN, Chesnutt JC, Chou L-S, et al. Detecting gait abnormalities after concussion or mild traumatic brain injury: a systematic review of single-task, dual-task, and complex gait. Gait & Posture. 2018;62:157–66.
140. Haider MN, Johnson SL, Mannix R, Macfarlane AJ, Constantino D, Johnson BD, et al. The Buffalo Concussion Bike Test for concussion assessment in adolescents. Sports health. 2019;11(6):492–7.
141. Blake TA, McKay CD, Meeuwisse WH, Emery CA. The impact of concussion on cardiac autonomic function: a systematic review. Brain injury. 2016;30(2):132–45.
142. Broglio SP, Williams R, Lapointe A, Rettmann A, Moore B, Meehan SK, et al. Brain Network Activation Technology Does Not Assist with Concussion Diagnosis and Return to Play in Football Athletes. Frontiers in Neurology. 2017;8.

143. Yue JK, Phelps RRL, Chandra A, Winkler EA, Manley GT, Berger MS. Sideline Concussion Assessment: The Current State of the Art. Neurosurgery. 2020;87(3):466–75.
144. Johnston W, Heiderscheit B, Coughlan G, McLoughlin R, Liston M, Brooks MA, et al. Concussion recovery evaluation using the inertial sensor instrumented Y Balance Test. Journal of neurotrauma. 2020;37(23):2549–57.
145. Doherty C, Zhao L, Ryan J, Komaba Y, Inomata A, Caulfield B. Quantification of postural control deficits in patients with recent concussion: an inertial-sensor based approach. Clinical biomechanics. 2017;42:79–84.
146. Pryhoda MK, Shelburne KB, Gorgens K, Ledreux A, Granholm A-C, Davidson BS. Centre of pressure velocity shows impairments in NCAA Division I athletes six months post-concussion during standing balance. Journal of sports sciences. 2020;38(23):2677–87.
147. Chin EY, Nelson LD, Barr WB, McCrory P, McCrea MA. Reliability and validity of the Sport Concussion Assessment Tool–3 (SCAT3) in high school and collegiate athletes. The American journal of sports medicine. 2016;44(9):2276–85.
148. Lanska DJ, editor The Romberg sign and early instruments for measuring postural sway. Seminars in neurology; 2002: Copyright© 2002 by Thieme Medical Publishers, Inc., 333 Seventh Avenue, New ….
149. Di Pietro V, Porto E, Ragusa M, Barbagallo C, Davies D, Forcione M, et al. Salivary MicroRNAs: diagnostic markers of mild traumatic brain injury in contact-sport. Frontiers in molecular neuroscience. 2018;11:290.
150. Di Pietro V, O'Halloran P, Watson CN, Begum G, Acharjee A, Yakoub KM, et al. Unique diagnostic signatures of concussion in the saliva of male athletes: the Study of Concussion in Rugby Union through MicroRNAs (SCRUM). British journal of sports medicine. 2021.
151. Howell DR, Lugade V, Potter MN, Walker G, Wilson JC. A multifaceted and clinically viable paradigm to quantify postural control impairments among adolescents with concussion. Physiological Measurement. 2019;40(8).
152. Swanenburg J, De Bruin ED, Uebelhart D, Mulder T. Falls prediction in elderly people: A 1-year prospective study. Gait & Posture. 2010;31(3):317–21.
153. Mansfield A, Inness EL. Force Plate Assessment of Quiet Standing Balance Control: Perspectives on Clinical Application within Stroke Rehabilitation. Rehabilitation Process and Outcome. 2015;4:RPO.S20363.
154. Howell DR, Osternig LR, Chou L-S. Consistency and cost of dual-task gait balance measure in healthy adolescents and young adults. Gait & Posture. 2016;49:176–80.
155. Menz HB, Latt MD, Tiedemann A, San Kwan MM, Lord SR. Reliability of the GAITRite® walkway system for the quantification of temporo-spatial parameters of gait in young and older people. Gait & posture. 2004;20(1):20–5.
156. Oldham JR, Difabio MS, Kaminski TW, Dewolf RM, Howell DR, Buckley TA. Efficacy of Tandem Gait to Identify Impaired Postural Control after Concussion. Medicine and science in sports and exercise. 2018;50(6):1162–8.
157. Hill MW, Wdowski MM, Rosicka K, Kay AD, Muehlbauer T. Exploring the relationship of static and dynamic balance with muscle mechanical properties of the lower limbs in healthy young adults. Frontiers in Physiology. 2023;14:881.
158. Morasso P. Integrating ankle and hip strategies for the stabilization of upright standing: An intermittent control model. Frontiers in Computational Neuroscience. 2022;16:956932.
159. Lynall RC, Campbell KR, Mauntel TC, Blackburn JT, Mihalik JP. Single-legged hop and single-legged squat balance performance in recreational athletes with a history of concussion. Journal of athletic training. 2020;55(5):488–93.
160. Reilly T, Morris T, Whyte G. The specificity of training prescription and physiological assessment: A review. Journal of Sports Sciences. 2009;27(6):575–89.

161. Twomey N, Diethe T, Fafoutis X, Elsts A, McConville R, Flach P, et al., editors. A comprehensive study of activity recognition using accelerometers. Informatics; 2018: Multidisciplinary Digital Publishing Institute.
162. Horsley BJ, Tofari PJ, Halson SL, Kemp JG, Dickson J, Maniar N, et al. Does Site Matter? Impact of Inertial Measurement Unit Placement on the Validity and Reliability of Stride Variables During Running: A Systematic Review and Meta-analysis. Sports Medicine. 2021:1–41.
163. Napier C, Willy RW, Hannigan BC, McCann R, Menon C. The effect of footwear, running speed, and location on the validity of two commercially available inertial measurement units during running. Frontiers in Sports and Active Living. 2021;3:102.
164. Boyd LJ, Ball K, Aughey RJ. The Reliability of MinimaxX Accelerometers for Measuring Physical Activity in Australian Football. International Journal of Sports Physiology and Performance. 2011;6(3):311–21.
165. Bredt SDGT, Chagas MH, Peixoto GH, Menzel HJ, Andrade AGPD. Understanding Player Load: Meanings and Limitations. Journal of Human Kinetics. 2020;71(1):5–9.
166. Castillo D, Weston M, McLaren SJ, Cámara J, Yanci J. Relationships Between Internal and External Match-Load Indicators in Soccer Match Officials. International Journal of Sports Physiology and Performance. 2017;12(7):922–7.
167. Barrett S, Midgley A, Lovell R. PlayerLoad™: Reliability, Convergent Validity, and Influence of Unit Position during Treadmill Running. International Journal of Sports Physiology and Performance. 2014;9(6):945–52.
168. Grafton ST, Ralston AB, Ralston JD. <p>Monitoring of postural sway with a head-mounted wearable device: effects of gender, participant state, and concussion</p>. Medical Devices: Evidence and Research. 2019;Volume 12:151–64.
169. Pitt W, Chen S-H, Chou L-S. Using IMU-based kinematic markers to monitor dual-task gait balance control recovery in acutely concussed individuals. Clinical Biomechanics. 2020;80:105145.
170. Crang ZL, Duthie G, Cole MH, Weakley J, Hewitt A, Johnston RD. The Validity and Reliability of Wearable Microtechnology for Intermittent Team Sports: A Systematic Review. Sports Medicine. 2020:1–17.
171. Cormack SJ, Mooney MG, Morgan W, McGuigan MR. Influence of neuromuscular fatigue on accelerometer load in elite Australian football players. International journal of sports physiology and performance. 2013;8(4):373–8.
172. Buchheit M, Lacome M, Cholley Y, Simpson BM. Neuromuscular Responses to Conditioned Soccer Sessions Assessed via GPS-Embedded Accelerometers: Insights Into Tactical Periodization. International Journal of Sports Physiology and Performance. 2018;13(5):577–83.
173. Leduc C, Tee J, Lacome M, Weakley J, Cheradame J, Ramirez C, et al. Convergent Validity, Reliability, and Sensitivity of a Running Test to Monitor Neuromuscular Fatigue. International Journal of Sports Physiology and Performance. 2020;15(8):1067–73.
174. Garrett J, Graham SR, Eston RG, Burgess DJ, Garrett LJ, Jakeman J, et al. A Novel Method of Assessment for Monitoring Neuromuscular Fatigue in Australian Rules Football Players. International Journal of Sports Physiology and Performance. 2019;14(5):598–605.
175. Rowell AE, Aughey RJ, Clubb J, Cormack SJ. A standardized small sided game can be used to monitor neuromuscular fatigue in professional A-league football players. Frontiers in physiology. 2018;9:1011.
176. Akenhead R, Marques JB, Paul DJ. Accelerometer load: a new way to measure fatigue during repeated sprint training? Science and Medicine in Football. 2017;1(2):151–6.
177. Fitzpatrick JF, Akenhead R, Russell M, Hicks KM, Hayes PR. Sensitivity and reproducibility of a fatigue response in elite youth football players. Science and Medicine in Football. 2019;3(3):214–20.
178. Johnston W, O'Reilly M, Argent R, Caulfield B. Reliability, Validity and Utility of Inertial Sensor Systems for Postural Control Assessment in Sport Science and Medicine Applications: A Systematic Review. Sports Medicine. 2019;49(5):783–818.

179. Horsley BJ, Tofari PJ, Halson SL, Kemp JG, Johnston RD, Cormack SJ. Thoracic-Worn Accelerometers Detect Fatigue-Related Changes in Vertical Stiffness During Sprinting. The Journal of Strength & Conditioning Research. 2022:10.1519.
180. Currell K, Jeukendrup AE. Validity, reliability and sensitivity of measures of sporting performance. Sports medicine. 2008;38(4):297–316.
181. Weir JP. Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. The Journal of Strength & Conditioning Research. 2005;19(1):231–40.
182. Kottner J, Streiner DL. The difference between reliability and agreement. Journal of clinical epidemiology. 2011;64(6):701.
183. Iverson GL, Gardner AJ. Incidence of Concussion and Time to Return-to-Play in the National Rugby League. Clinical journal of sport medicine. 2022;32(6):595–9.
184. Lindberg MA, Moy Martin EM, Marion DW. Military Traumatic Brain Injury: The History, Impact, and Future. Journal of neurotrauma. 2022;39(17-18):1133–45.
185. Walton SR, Malin SK, Kranz S, Broshek DK, Hertel J, Resch JE. Whole-body metabolism, carbohydrate utilization, and caloric energy balance after sport concussion: a pilot study. Sports health. 2020;12(4):382–9.
186. Purcell L, Harvey J, Seabrook JA. Patterns of Recovery Following Sport-Related Concussion in Children and Adolescents. Clin Pediatr (Phila). 2016;55(5):452–8.
187. Echemendia RJ, Meeuwisse W, McCrory P, Davis GA, Putukian M, Leddy J, et al. The Sport Concussion Assessment Tool 5th Edition (SCAT5). British Journal of Sports Medicine. 2017:bjsports–2017–0.
188. Lynall RC, Mauntel TC, Pohlig RT, Kerr ZY, Dompier TP, Hall EE, et al. Lower extremity musculoskeletal injury risk after concussion recovery in high school athletes. Journal of athletic training. 2017;52(11):1028–34.
189. Johnston W, Coughlan GF, Caulfield B. Challenging concussed athletes: the future of balance assessment in concussion. Qjm-an International Journal of Medicine. 2017;110(12):779–83.
190. Salisbury JP, Keshav NU, Sossong AD, Sahin NT. Concussion assessment with smartglasses: Validation study of balance measurement toward a lightweight, multimodal, field-ready platform. Journal of Medical Internet Research. 2018;20(1).
191. Page MJ, McKenzie JE, Bossuyt PM, Boutron I, Hoffmann TC, Mulrow CD, et al. The PRISMA 2020 statement: an updated guideline for reporting systematic reviews. Systematic Reviews. 2021;10(1).
192. Wright JG, Swiontkowski MF, Heckman JD. Introducing Levels of Evidence to The Journal. JBJS. 2003;85(1):1–3.
193. Mokkink LB, De Vet HCW, Prinsen CAC, Patrick DL, Alonso J, Bouter LM, et al. COSMIN Risk of Bias checklist for systematic reviews of Patient-Reported Outcome Measures. Quality of Life Research. 2018;27(5):1171–9.
194. Mokkink L, Boers M, Vleuten C, Patrick D, Alonso J, Bouter L, et al. COSMIN Risk of Bias tool to assess the quality of studies on reliability and measurement error of outcome measurement instrument. 2021.
195. Prinsen CAC, Mokkink LB, Bouter LM, Alonso J, Patrick DL, De Vet HCW, et al. COSMIN guideline for systematic reviews of patient-reported outcome measures. Quality of Life Research. 2018;27(5):1147–57.
196. Hearn MC, Levy SS, Baweja HS, Goble DJ. BTrackS Balance Test for Concussion Management is Resistant to Practice Effects. Clinical Journal of Sport Medicine. 2018;28(2):177–9.
197. Chang JO, Levy SS, Seay SW, Goble DJ. An alternative to the balance error scoring system: using a low-cost balance board to improve the validity/reliability of sports-related concussion balance testing. Clinical journal of sport medicine : official journal of the Canadian Academy of Sport Medicine. 2014;24(3):256–62.
198. Hunt TN, Ferrara MS, Bornstein RA, Baumgartner TA. The reliability of the modified balance error scoring system. Clinical Journal of Sport Medicine. 2009;19(6):471–5.

199. Broglio SP, Zhu W, Sopiarz K, Park Y. Generalizability Theory Analysis of Balance Error Scoring System Reliability in Healthy Young Adults. Journal of Athletic Training. 2009;44(5):497–502.
200. Johnston W, Martin, Garrett, Caulfield B. Inertial Sensor Technology Can Capture Changes in Dynamic Balance Control during the Y Balance Test. Digital Biomarkers. 2018;1(2):106–17.
201. Lininger MR, Leahy TE, Haug EC, Bowman TG. TEST-RETEST RELIABILITY OF THE LIMITS OF STABILITY TEST PERFORMED BY YOUNG ADULTS USING NEUROCOM (R) VSR SPORT. International Journal of Sports Physical Therapy. 2018;13(5):800–7.
202. Finnoff JT, Peterson VJ, Hollman JH, Smith J. Intrarater and interrater reliability of the Balance Error Scoring System (BESS). Pm&r. 2009;1(1):50–4.
203. Riemann BL, Guskiewicz KM, Shields EW. Relationship between Clinical and Forceplate Measures of Postural Stability. Journal of Sport Rehabilitation. 1999;8(2):71–82.
204. Kleffelgaard I, Langhammer B, Sandhaug M, Pripp AH, Søberg HL. Measurement properties of the modified and total Balance ERror SCoring System–the BESS, in a healthy adult sample. European journal of Physiotherapy. 2018;20(1):25–31.
205. Glass SM, Napoli A, Thompson ED, Obeid I, Tucker CA. Validity of an automated balance error scoring system. Journal of Applied Biomechanics. 2019;35(1):32–6.
206. Prim JH, Favorov OV, Cecchini AS, Scherer MR, Weightman MM, McCulloch KL. Clinical Utility and Analysis of the Run-Roll-Aim Task: Informing Return-to-Duty Readiness Decisions in Active-Duty Service Members. Military medicine. 2019;184(5-6):e268–e77.
207. Buckley TA, Munkasy BA, Clouse BP. Sensitivity and specificity of the modified balance error scoring system in concussed collegiate student athletes. Clinical journal of sport medicine. 2018;28(2):174–6.
208. King LA, Horak FB, Mancini M, Pierce D, Priest KC, Chesnutt J, et al. Instrumenting the balance error scoring system for use with patients reporting persistent balance problems after mild traumatic brain injury. Archives of physical medicine and rehabilitation. 2014;95(2):353–9.
209. Ralston JD, Raina A, Benson BW, Peters RM, Roper JM, Ralston AB. Physiological Vibration Acceleration (Phybrata) Sensor Assessment of Multi-System Physiological Impairments and Sensory Reweighting Following Concussion. Medical devices (Auckland, NZ). 2020;13:411–38.
210. Teel EF, Gay MR, Arnett PA, Slobounov SM. Differential sensitivity between a virtual reality balance module and clinically used concussion balance modalities. Clinical Journal of Sport Medicine. 2016;26(2):162–6.
211. Fyffe A, Bogg T, Orr R, Browne GJ. Association of Simple Step Test With Readiness for Exercise in Youth After Concussion. Journal of Head Trauma Rehabilitation. 2020;35(2):E95–E102.
212. Favorov O, Kursun O, Challener T, Cecchini A, McCulloch KL. Wearable Sensors Detect Movement Differences in the Portable Warrior Test of Tactical Agility After mTBI in Service Members. Military medicine. 2021.
213. Broglio SP, Katz BP, Zhao S, McCrea M, McAllister T. Test-Retest Reliability and Interpretation of Common Concussion Assessment Tools: Findings from the NCAA-DoD CARE Consortium. Sports medicine (Auckland, NZ). 2018;48(5):1255–68.
214. Alsalaheen BA, Haines J, Yorke A, Stockdale K, P. Broglio S. Reliability and concurrent validity of instrumented balance error scoring system using a portable force plate system. The Physician and Sportsmedicine. 2015;43(3):221–6.
215. Nelson LD, Loman MM, LaRoche AA, Furger RE, McCrea MA. Baseline Performance and Psychometric Properties of the Child Sport Concussion Assessment Tool 3 (Child-SCAT3) in 5- to 13-year-old Athletes. Clinical journal of sport medicine : official journal of the Canadian Academy of Sport Medicine. 2017;27(4):381–7.
216. Kontos AP, Monti K, Eagle SR, Thomasma E, Holland CL, Thomas D, et al. Test-retest reliability of the Vestibular Ocular Motor Screening (VOMS) tool and modified Balance Error Scoring System (mBESS) in US military personnel. Journal of Science & Medicine in Sport. 2021;24(3):264–8.

217. Hänninen T, Parkkari J, Tuominen M, Öhman J, Howell DR, Iverson GL, et al. Sport Concussion Assessment Tool: Interpreting day-of-injury scores in professional ice hockey players. Journal of Science & Medicine in Sport. 2018;21(8):794–9.
218. Corwin DJ, McDonald CC, Arbogast KB, Mohammed FN, Metzger KB, Pfeiffer MR, et al. Clinical and Device-based Metrics of Gait and Balance in Diagnosing Youth Concussion. Medicine and Science in Sports and Exercise. 2020;52(3):542–8.
219. King LA, Mancini M, Fino PC, Chesnutt J, Swanson CW, Markwardt S, et al. Sensor-Based Balance Measures Outperform Modified Balance Error Scoring System in Identifying Acute Concussion. Annals of Biomedical Engineering. 2017;45(9):2135–45.
220. Schneiders AG, Sullivan SJ, McCrory PR, Gray A, Maruthayanar S, Singh P, et al. The effect of exercise on motor performance tasks used in the neurological assessment of sports-related concussion. British Journal of Sports Medicine. 2008;42(12):1011–3.
221. Schneiders AG, Sullivan SJ, Gray AR, Hammond-Tooke GD, McCrory PR. Normative values for three clinical measures of motor performance used in the neurological assessment of sports concussion. Journal of Science and Medicine in Sport. 2010;13(2):196–201.
222. German D, Bahat HS. Validity and Reliability of a Customized Smartphone Application for Postural Sway Assessment. Journal of Manipulative and Physiological Therapeutics. 2021;44(9):707–17.
223. Baracks J, Casa DJ, Covassin T, Sacko R, Scarneo SE, Schnyer D, et al. Acute sport-related concussion screening for collegiate athletes using an instrumented balance assessment. Journal of athletic training. 2018;53(6):597–605.
224. Westwood C, Killelea C, Faherty M, Sell T. Postural Stability Under Dual-Task Conditions: Development of a Post-Concussion Assessment for Lower-Extremity Injury Risk. Journal of Sport Rehabilitation. 2020;29(1):131–3.
225. Marchetti GF, Bellanca J, Whitney SL, Lin JC-C, Musolino MC, Furman GR, et al. The development of an accelerometer-based measure of human upright static anterior-posterior postural sway under various sensory conditions: Test–retest reliability, scoring and preliminary validity of the Balance Accelerometry Measure (BAM). Journal of Vestibular Research. 2013;23(4-5):227–35.
226. Furman GR, Lin CC, Bellanca JL, Marchetti GF, Collins MW, Whitney SL. Comparison of the balance accelerometer measure and balance error scoring system in adolescent concussions in sports. American Journal of Sports Medicine. 2013;41(6):1404–10.
227. Goble DJ, Manyak KA, Abdenour TE, Rauh MJ, Baweja HS. An initial evaluation of the BTrackS balance plate and sports balance software for concussion diagnosis. International Journal of Sports Physical Therapy. 2016;11(2):149.
228. Amick RZ, Chaparro A, Patterson JA, Jorgensen MJ. Test-retest reliability of the sway balance mobile application. Journal of mobile technology in medicine. 2015;4(2):40.
229. Register-Mihalik JK, Guskiewicz KM, Mihalik JP, Schmidt JD, Kerr ZY, McCrea MA. Reliable change, sensitivity, and specificity of a multidimensional concussion assessment battery: implications for caution in clinical practice. The Journal of head trauma rehabilitation. 2013;28(4):274–83.
230. Resch JE, Brown CN, Schmidt J, Macciocchi SN, Blueitt D, Cullum CM, et al. The sensitivity and specificity of clinical measures of sport concussion: three tests are better than one. BMJ Open Sport & Exercise Medicine. 2016;2(1):e000012.
231. Christy JB, Cochrane GD, Almutairi A, Busettini C, Swanson MW, Weise KK. Peripheral Vestibular and Balance Function in Athletes With and Without Concussion. Journal of Neurologic Physical Therapy. 2019;43(3):153–9.
232. Broglio SP, Ferrara MS, Sopiarz K, Kelly MS. Reliable Change of the Sensory Organization Test. Clinical Journal of Sport Medicine. 2008;18(2):148–54.

233. Toong T, Wilson KE, Hunt AW, Scratch S, DeMatteo C, Reed N. Sensitivity and Specificity of a Multimodal Approach for Concussion Assessment in Youth Athletes. Journal of sport rehabilitation. 2021:1–10.
234. Teel EF, Slobounov SM. Validation of a Virtual Reality Balance Module for Use in Clinical Concussion Assessment and Management. Clinical Journal of Sport Medicine. 2015;25(2):144–8.
235. Stephens JA, Davies PL, Gavin WJ, Mostofsky SH, Slomine BS, Suskauer SJ. Evaluating Motor Control Improves Discrimination of Adolescents with and without Sports Related Concussion. Journal of Motor Behavior. 2020;52(1):13–21.
236. Johnston W, O'Reilly M, Duignan C, Liston M, McLoughlin R, Coughlan GF, et al. Association of Dynamic Balance With Sports-Related Concussion: A Prospective Cohort Study. American Journal of Sports Medicine. 2019;47(1):197–205.
237. Alsalaheen B, Haines J, Yorke A, Broglio SP. Reliability and Construct Validity of Limits of Stability Test in Adolescents Using a Portable Forceplate System. Archives of Physical Medicine and Rehabilitation. 2015;96(12):2194–200.
238. Gagnon I, Swaine B, Friedman D, Forget R. Children show decreased dynamic balance after mild traumatic brain injury. Archives of Physical Medicine and Rehabilitation. 2004;85(3):444–52.
239. Rao HM, Talkar T, Ciccarelli G, Nolan M, O'Brien A, Vergara-Diaz G, et al. Sensorimotor conflict tests in an immersive virtual environment reveal subclinical impairments in mild traumatic brain injury. Scientific Reports. 2020;10(1).
240. Pape MM, Williams K, Kodosky PN, Dretsch M. The Community Balance and Mobility Scale: A Pilot Study Detecting Impairments in Military Service Members With Comorbid Mild TBI and Psychological Health Conditions. Journal of Head Trauma Rehabilitation. 2016;31(5):339–45.
241. Pape MM, Kodosky PN, Hoover P. The Community Balance and Mobility Scale: Detecting Impairments in Military Service Members With Mild Traumatic Brain Injury. Military Medicine. 2020;185(3-4):428–35.
242. Eemanipure S, Shafinia P, Shabani SEHS, Ghotbi-Varzaneh A. Identify normative values of balance tests toward neurological assessment of sports related concussion. Iranian Rehabilitation Journal. 2012;10(15):39–43.
243. Howell DR, Brilliant AN, Meehan WP. Tandem Gait Test-Retest Reliability Among Healthy Child and Adolescent Athletes. Journal of Athletic Training. 2019;54(12):1254–9.
244. Wingerson MJ, Seehusen CN, Walker G, Wilson JC, Howell DR. Clinical feasibility and utility of a dual-task tandem gait protocol for pediatric concussion management. Journal of athletic training. 2020.
245. Barnes A, Smulligan K, Wingerson MJ, Little C, Lugade V, Wilson JC, et al. A multifaceted approach to interpreting reaction time deficits after adolescent concussion. Journal of athletic training. 2023.
246. Howell DR, Lugade V, Taksir M, Meehan WP. Determining the utility of a smartphone-based gait evaluation for possible use in concussion management. Physician and Sportsmedicine. 2020;48(1):75–80.
247. Howell DR, Oldham JR, DiFabio M, Vallabhajosula S, Hall EE, Ketcham CJ, et al. Single-task and dual-task gait among collegiate athletes of different sport classifications: Implications for concussion management. Journal of Applied Biomechanics. 2017;33(1):24–31.
248. Nishiguchi S, Yamada M, Nagai K, Mori S, Kajiwara Y, Sonoda T, et al. Reliability and Validity of Gait Analysis by Android-Based Smartphone. Telemedicine and e-Health. 2012;18(4):292–6.
249. Silsupadol P, Teja K, Lugade V. Reliability and validity of a smartphone-based assessment of gait parameters across walking speed and smartphone locations: Body, bag, belt, hand, and pocket. Gait & posture. 2017;58:516–22.
250. Kuznetsov NA, Robins RK, Long B, Jakiela JT, Haran FJ, Ross SE, et al. Validity and reliability of smartphone orientation measurement to quantify dynamic balance function. Physiological measurement. 2018;39(2):02NT1.

251. Howell DR, Seehusen CN, Wingerson MJ, Wilson JC, Lynall RC, Lugade V. Reliability and minimal detectable change for a smartphone-based motor-cognitive assessment: Implications for concussion management. Journal of Applied Biomechanics. 2021;37(4):380–7.
252. Wilkerson GB, Nabhan DC, Perry TS. A Novel Approach to Assessment of Perceptual-Motor Efficiency and Training-Induced Improvement in the Performance Capabilities of Elite Athletes. Frontiers in Sports and Active Living. 2021;3.
253. Howell D, Osternig L, Chou LS. Monitoring recovery of gait balance control following concussion using an accelerometer. Journal of Biomechanics. 2015;48(12):3364–8.
254. Cecchini AS, Prim J, Zhang W, Harrison CH, McCulloch KL. The Portable Warrior Test of Tactical Agility: A Novel Functional Assessment That Discriminates Service Members Diagnosed With Concussion From Controls. Military medicine. 2021.
255. Bell DR, Guskiewicz KM, Clark MA, Padua DA. Systematic review of the balance error scoring system. Sports health. 2011;3(3):287–95.
256. Ulman S, Erdman AL, Loewen A, Worrall HM, Tulchin-Francis K, Jones JC, et al. Improvement in balance from diagnosis to return-to-play initiation following a sport-related concussion: BESS scores vs center-of-pressure measures. Brain injury. 2022;36(8):921–30.
257. McCrea M, Guskiewicz K, Marshall S, Barr W, Randolph C, Cantu R. Acute effects and recovery time following concussion in collegiate football players. Sports Med Update. 2004;38:369–71.
258. McCulloch KL, Buxton E, Hackney J, Lowers S. Balance, Attention, and Dual-Task Performance During Walking After Brain Injury: Associations With Falls History. The Journal of Head Trauma Rehabilitation. 2010;25(3):155–63.
259. Nordström A, Nordström P, Ekstrand J. Sports-related concussion increases the risk of subsequent injury by about 50% in elite male football players. British Journal of Sports Medicine. 2014;48(19):1447–50.
260. Australian Government ASC. Participation across life stages In: Sport Australia A, editor. 2020.
261. Veale JP, Pearce AJ, Buttifant D, Carlson JS. Anthropometric profiling of elite junior and senior Australian football players. International journal of sports physiology and performance. 2010;5(4):509–20.
262. Fuller CW, Taylor AE, Brooks JH, Kemp SP. Changes in the stature, body mass and age of English professional rugby players: a 10-year review. Journal of Sports Sciences. 2013;31(7):795–802.
263. Roberts SP, Trewartha G, England M, Shaddick G, Stokes KA. Epidemiology of time-loss injuries in English community-level rugby union. BMJ open. 2013;3(11):e003998.
264. Lovell MR, Collins MW. Neuropsychological assessment of the college football player. The journal of head trauma rehabilitation. 1998;13(2):9–26.
265. Crang ZL, Duthie G, Cole MH, Weakley J, Hewitt A, Johnston RD. The inter-device reliability of global navigation satellite systems during team sport movement across multiple days. Journal of science and medicine in sport. 2022;25(4):340–4.
266. Catena RD, van Donkelaar P, Chou L-S. Altered balance control following concussion is better detected with an attention test during gait. Gait & posture. 2007;25(3):406–11.
267. Parker TM, Osternig LR, Lee H-J, Donkelaar PV, Chou L-S. The effect of divided attention on gait stability following concussion. Clinical Biomechanics. 2005;20(4):389–95.
268. Powell D, Stuart S, Godfrey A. Exploring Inertial-Based Wearable Technologies for Objective Monitoring in Sports-Related Concussion: A Single-Participant Report. Physical Therapy. 2022;102(5).
269. Catena RD. Attention and gait performance following a concussion: University of Oregon; 2008.
270. Ayala F, Calderón-López A, Delgado-Gosálbez JC, Parra-Sánchez S, Pomares-Noguera C, Hernández-Sánchez S, et al. Acute Effects of Three Neuromuscular Warm-Up Strategies on Several Physical Performance Measures in Football Players. PLOS ONE. 2017;12(1):e0169660.

271. Williams RD, Gillham S, Highton J, Twist C. The influence of warm-up duration on simulated rugby league interchange match performance. Science and Medicine in Football. 2021;5(2):137–43.
272. Thornton HR, Nelson AR, Delaney JA, Serpiello FR, Duthie GM. Interunit Reliability and Effect of Data-Processing Methods of Global Positioning Systems. International journal of sports physiology and performance. 2019;14(4):1–438.
273. Beato M, Coratella G, Stiff A, Iacono AD. The Validity and Between-Unit Variability of GNSS Units (STATSports Apex 10 and 18 Hz) for Measuring Distance and Peak Speed in Team Sports. Frontiers in Physiology. 2018;9.
274. Beato M, de Keijzer K. The inter-unit and inter-model reliability of GNSS STATSports Apex and Viper units in measuring peak speed over 5, 10, 15, 20 and 30 meters. Biology of Sport. 2019;36(4):317–21.
275. Scott MT, Scott TJ, Kelly VG. The validity and reliability of global positioning systems in team sport: a brief review. The Journal of Strength & Conditioning Research. 2016;30(5):1470–90.
276. Wundersitz DWT, Gastin PB, Richter C, Robertson SJ, Netto KJ. Validity of a trunk-mounted accelerometer to assess peak accelerations during walking, jogging and running. European Journal of Sport Science. 2015;15(5):382–90.
277. Wundersitz DW, Gastin P, Robertson S, Davey P, Netto K. Validation of a trunk-mounted accelerometer to measure peak impacts during team sport movements. International journal of sports medicine. 2015;36(09):742–6.
278. Berg-Hansen P, Moen SM, Austeng A, Gonzales V, Klyve TD, Negård H, et al. Sensor-based gait analyses of the six-minute walk test identify qualitative improvement in gait parameters of people with multiple sclerosis after rehabilitation. Journal of Neurology. 2022;269(7):3723–34.
279. Washabaugh EP, Kalyanaraman T, Adamczyk PG, Claflin ES, Krishnan C. Validity and repeatability of inertial measurement units for measuring gait parameters. Gait & Posture. 2017;55:87–93.
280. Zhou L, Fischer E, Tunca C, Brahms CM, Ersoy C, Granacher U, et al. How We Found Our IMU: Guidelines to IMU Selection and a Comparison of Seven IMUs for Pervasive Healthcare Applications. Sensors. 2020;20(15):4090.
281. Dugan EL, Shilt JS, Masterson CM, Ernest KM. The use of inertial measurement units to assess gait and postural control following concussion. Gait & Posture. 2021;83:262–7.
282. Morelli N, Heebner NR, Bergin RT, Quintana C, Hoch MC. The effect of cognitive dual-tasks on dynamic postural sway during gait using inertial measurement units. Physiological Measurement. 2020;41(4).
283. Wittmann F, Lambercy O, Gassert R. Magnetometer-Based Drift Correction During Rest in IMU Arm Motion Tracking. Sensors. 2019;19(6):1312.
284. Armitage M, Beato M, McErlain-Naylor SA. Inter-unit reliability of IMU Step metrics using IMeasureU Blue Trident inertial measurement units for running-based team sport tasks. Journal of sports sciences. 2021;39(13):1512–8.
285. Burland JP, Outerleys JB, Lattermann C, Davis IS. Reliability of wearable sensors to assess impact metrics during sport-specific tasks. Journal of Sports Sciences. 2021;39(4):406–11.
286. Dimmick HL, Van Rassel CR, Macinnis MJ, Ferber R. Between-Day Reliability of Commonly Used IMU Features during a Fatiguing Run and the Effect of Speed. Sensors. 2022;22(11):4129.
287. Delves RI, Aughey RJ, Ball K, Duthie GM. The quantification of acceleration events in elite team sport: a systematic review. Sports Medicine-Open. 2021;7(1):1–35.
288. Valenciano PJ, Conceicao NR, Marcori AJ, Teixeira LA. Use of accelerometry to investigate standing and dynamic body balance in people with cerebral palsy: A systematic review. Gait & Posture. 2022.
289. Day EM, Alcantara RS, McGeehan MA, Grabowski AM, Hahn ME. Low-pass filter cutoff frequency affects sacral-mounted inertial measurement unit estimations of peak vertical ground reaction force and contact time during treadmill running. Journal of Biomechanics. 2021;119:110323.

290. Horsley BJ, Tofari PJ, Halson SL, Kemp JG, Chalkley D, Cole MH, et al. Validity and reliability of thoracic-mounted inertial measurement units to derive gait characteristics during running. The Journal of Strength & Conditioning Research. 2022:10.1519.
291. Costa M, Peng C-K, Goldberger AL, Hausdorff JM. Multiscale entropy analysis of human gait dynamics. Physica A: Statistical Mechanics and its applications. 2003;330(1-2):53–60.
292. Yentes JM, Raffalt PC. Entropy analysis in gait research: methodological considerations and recommendations. Annals of biomedical engineering. 2021;49:979–90.
293. Yentes JM, Hunt N, Schmid KK, Kaipust JP, McGrath D, Stergiou N. The appropriate use of approximate entropy and sample entropy with short data sets. Annals of biomedical engineering. 2013;41:349–65.
294. Rabuffetti M, Scalera GM, Ferrarin M. Effects of gait strategy and speed on regularity of locomotion assessed in healthy subjects using a multi-sensor method. Sensors. 2019;19(3):513.
295. Sosnoff JJ, Broglio SP, Shin S, Ferrara MS. Previous mild traumatic brain injury and postural-control dynamics. Journal of athletic training. 2011;46(1):85–91.
296. Gao J, Hu J, Buckley T, White K, Hass C. Shannon and Renyi entropies to classify effects of mild traumatic brain injury on postural sway. PLoS One. 2011;6(9):e24446.
297. Cavanaugh JT, Guskiewicz KM, Giuliani C, Marshall S, Mercer VS, Stergiou N. Recovery of postural control after cerebral concussion: new insights using approximate entropy. Journal of athletic training. 2006;41(3):305.
298. Asghari M, Ehsani H, Cohen A, Tax T, Mohler J, Toosizadeh N. Nonlinear analysis of the movement variability structure can detect aging-related differences among cognitively healthy individuals. Human movement science. 2021;78:102807.
299. McCamley JD, Denton W, Arnold A, Raffalt PC, Yentes JM. On the calculation of sample entropy using continuous and discrete human gait data. Entropy. 2018;20(10):764.
300. Bonnette S, Diekfuss JA, Grooms D, Myer GD, Meehan III WP, Howell DR. Integrated linear and nonlinear trunk dynamics identify residual concussion deficits. Neuroscience letters. 2020;729:134975.
301. Ahmadi S, Sepehri N, Wu C, Szturm T. Sample entropy of human gait center of pressure displacement: a systematic methodological analysis. Entropy. 2018;20(8):579.
302. Randolph C, Millis S, Barr WB, McCrea M, Guskiewicz KM, Hammeke TA, et al. Concussion Symptom Inventory: An Empirically Derived Scale for Monitoring Resolution of Symptoms Following Sport-Related Concussion. Archives of Clinical Neuropsychology. 2009;24(3):219–29.
303. Merritt VC, Meyer JE, Arnett PA. A novel approach to classifying postconcussion symptoms: The application of a new framework to the Post-Concussion Symptom Scale. Journal of Clinical and Experimental Neuropsychology. 2015;37(7):764–75.
304. McClincy MP, Lovell MR, Pardini J, Collins MW, Spore MK. Recovery from sports concussion in high school and collegiate athletes. Brain Injury. 2006;20(1):33–9.
305. Field M, Collins MW, Lovell MR, Maroon J. Does age play a role in recovery from sports-related concussion? A comparison of high school and collegiate athletes. The Journal of pediatrics. 2003;142(5):546–53.
306. Covassin T, Elbin RJ, Harris W, Parker T, Kontos A. The Role of Age and Sex in Symptoms, Neurocognitive Performance, and Postural Stability in Athletes After Concussion. The American Journal of Sports Medicine. 2012;40(6):1303–12.
307. Lovell MR, Iverson GL, Collins MW, Podell K, Johnston KM, Pardini D, et al. Measurement of Symptoms Following Sports-Related Concussion: Reliability and Normative Data for the Post-Concussion Scale. Applied Neuropsychology. 2006;13(3):166–74.
308. Borg GA. Psychophysical bases of perceived exertion. Medicine & science in sports & exercise. 1982.
309. Williams N. The Borg Rating of Perceived Exertion (RPE) scale. Occupational Medicine. 2017;67(5):404–5.

310. Hinds A, Leddy J, Freitas M, Willer B. The Effect of Exertion on Heart Rate and Rating of Perceived Exertion in Acutely Concussed Individuals. Journal of Neurology & Neurophysiology. 2016;7(4).
311. Chen Y-L, Chen C-C, Hsia P-Y, Lin S-K. Relationships of Borg's Rpe 6–20 Scale and Heart Rate in Dynamic and Static Exercises among a Sample of Young Taiwanese Men. Perceptual and Motor Skills. 2013;117(3):971–82.
312. Cabral LL, Lopes PB, Wolf R, Stefanello JMF, Pereira G. A SYSTEMATIC REVIEW OF CROSS-CULTURAL ADAPTATION AND VALIDATION OF BORG'S RATING OF PERCEIVED EXERTION SCALE. Journal of Physical Education. 2017;28(1).
313. Serafim THS, Tognato AC, Nakamura PM, Queiroga MR, Pereira G, Nakamura FY, et al. Development of the Color Scale of Perceived Exertion: Preliminary Validation. Perceptual and Motor Skills. 2014;119(3):884–900.
314. Arney BE, Glover R, Fusco A, Cortis C, de Koning JJ, van Erp T, et al. Comparison of RPE (rating of perceived exertion) scales for session RPE. International journal of sports physiology and performance. 2019;14(7):994–6.
315. Chung P-K, Zhao Y, Liu J-D, Quach B. A Brief Note on the Validity and Reliability of the Rating of Perceived Exertion Scale in Monitoring Exercise Intensity among Chinese Older Adults in Hong Kong. Perceptual and Motor Skills. 2015;121(3):805–9.
316. Karavatas SG, Tavakol K. Concurrent validity of Borg's rating of perceived exertion in african-american young adults, employing heart rate as the standard. Internet Journal of Allied Health Sciences and Practice. 2005;3(1):5.
317. Wenos D, Wallace J, Surburg P, Morris H. Reliability and comparison of RPE during variable and constant exercise protocols performed by older women. International journal of sports medicine. 1996;17(03):193–8.
318. Mahon A, Marsh M. Reliability of the rating of perceived exertion at ventilatory threshold in children. International journal of sports medicine. 1992;13(08):567–71.
319. Kuo AD, Donelan JM. Dynamic principles of gait and their clinical implications. Physical therapy. 2010;90(2):157–74.
320. Macie A, Matson T, Schinkel-Ivy A. Age affects the relationships between kinematics and postural stability during gait. Gait & Posture. 2023;102:86–92.
321. Creaby MW, Cole MH. Gait characteristics and falls in Parkinson's disease: A systematic review and meta-analysis. Parkinsonism & related disorders. 2018;57:1–8.
322. Brodie MA, Coppens MJ, Ejupi A, Gschwind YJ, Annegarn J, Schoene D, et al. Comparison between clinical gait and daily-life gait assessments of fall risk in older people. Geriatrics & gerontology international. 2017;17(11):2274–82.
323. Verghese J, Holtzer R, Lipton RB, Wang C. Quantitative gait markers and incident fall risk in older adults. Journals of Gerontology Series A: Biomedical Sciences and Medical Sciences. 2009;64(8):896–901.
324. Hausdorff JM, Rios DA, Edelberg HK. Gait variability and fall risk in community-living older adults: a 1-year prospective study. Archives of physical medicine and rehabilitation. 2001;82(8):1050–6.
325. Cole MH, Silburn PA, Wood JM, Worringham CJ, Kerr GK. Falls in Parkinson's disease: kinematic evidence for impaired head and trunk control. Movement Disorders. 2010;25(14):2369–78.
326. Wood JM, Lacherez PF, Black AA, Cole MH, Boon MY, Kerr GK. Postural stability and gait among older adults with age-related maculopathy. Investigative ophthalmology & visual science. 2009;50(1):482–7.
327. Prasanth H, Caban M, Keller U, Courtine G, Ijspeert A, Vallery H, et al. Wearable sensor-based real-time gait detection: A systematic review. Sensors. 2021;21(8):2727.

328. Thevathasan W, Cole MH, Graepel CL, Hyam JA, Jenkinson N, Brittain J-S, et al. A spatiotemporal analysis of gait freezing and the impact of pedunculopontine nucleus stimulation. Brain. 2012;135(5):1446–54.
329. Fusca M, Negrini F, Perego P, Magoni L, Molteni F, Andreoni G. Validation of a wearable IMU system for gait analysis: Protocol and application to a new system. Applied Sciences. 2018;8(7):1167.
330. Lyle MA, Jensen JC, Hunnicutt JL, Brown JJ, Chambliss CP, Newsome MA, et al. Associations of strength and spatiotemporal gait variables with knee loading during gait after anterior cruciate ligament reconstruction. Journal of Athletic Training. 2022;57(2):158–64.
331. Hadizadeh M, Amri S, Roohi S, Mohafez H. Assessment of gait symmetry improvements in national athletes after anterior cruciate ligament reconstruction during rehabilitation. International journal of sports medicine. 2016:997–1002.
332. Previs KM. Changes in Gait Characteristics Following a Concussion. 2012.
333. Kieffer EE, Brolinson PG, Rowson S. Dual-task gait performance following head impact exposure in male and female collegiate rugby players. International Journal of Sports Physical Therapy. 2022;17(3):355.
334. Ross LK, Lalka A, Keeter C, Hamati MC, Robell K, Howell DR, et al. Playing surface and sport contact status modulate time to lower extremity musculoskeletal injury at a greater level than concussion history among collegiate student-athletes. Journal of ISAKOS. 2024.
335. Howell D, Osternig L, Chou L-S. Monitoring recovery of gait balance control following concussion using an accelerometer. Journal of Biomechanics. 2015;48(12):3364–8.
336. Patricios JS, Davis GA, Ahmed OH, Blauwet C, Schneider GM, Purcell LK, et al. Introducing the sport concussion office assessment tool 6 (SCOAT6). BMJ Publishing Group Ltd and British Association of Sport and Exercise Medicine; 2023. p. 648–50.
337. Yue JK, Phelps RR, Chandra A, Winkler EA, Manley GT, Berger MS. Sideline concussion assessment: the current state of the art. Neurosurgery. 2020;87(3):466–75.
338. Cunningham J, Broglio S, Wyse J, Mc Hugh C, Farrell G, Denvir K, et al. Athlete concussion history recall is underestimated: A validation study of self-reported concussion history among current professional rugby union players. Brain injury. 2021;35(1):65–71.
339. Zou G. Sample size formulas for estimating intraclass correlation coefficients with precision and assurance. Statistics in medicine. 2012;31(29):3972–81.
340. Borg G. The perception of physical performance. Frontiers of fitness. 1971:280–94.
341. Delgado-Bonal A, Marshak A. Approximate entropy and sample entropy: A comprehensive tutorial. Entropy. 2019;21(6):541.
342. Koo TK, Li MY. A Guideline of Selecting and Reporting Intraclass Correlation Coefficients for Reliability Research. Journal of Chiropractic Medicine. 2016;15(2):155–63.
343. Sainani KL. Reliability statistics. PM&R. 2017;9(6):622–8.
344. Ryan NS, Bruno PA, Barden JM. Test–Retest Reliability and the Effects of Walking Speed on Stride Time Variability During Continuous, Overground Walking in Healthy Young Adults. Journal of Applied Biomechanics. 2020;37(2):102–8.
345. Fujiwara S, Sato S, Sugawara A, Nishikawa Y, Koji T, Nishimura Y, et al. The coefficient of variation of step time can overestimate gait abnormality: Test-retest reliability of gait-related parameters obtained with a tri-axial accelerometer in healthy subjects. Sensors. 2020;20(3):577.
346. Thies SB, Richardson JK, Ashton-Miller JA. Effects of surface irregularity and lighting on step variability during gait:: A study in healthy young and older women. Gait & posture. 2005;22(1):26–31.
347. Nohelova D, Bizovska L, Vuillerme N, Svoboda Z. Gait variability and complexity during single and dual-task walking on different surfaces in outdoor environment. Sensors. 2021;21(14):4792.
348. Karas M, Bai J, Strączkiewicz M, Harezlak J, Glynn NW, Harris T, et al. Accelerometry data in health research: challenges and opportunities: review and examples. Statistics in biosciences. 2019;11:210–37.

349. Collins SH, Kuo AD. Two independent contributions to step variability during over-ground human walking. PloS one. 2013;8(8):e73597.
350. König N, Singh NB, Von Beckerath J, Janke L, Taylor WR. Is gait variability reliable? An assessment of spatio-temporal parameters of gait variability during continuous overground walking. Gait & posture. 2014;39(1):615–7.
351. Paraschiv-Ionescu A, Newman CJ, Carcreff L, Gerber CN, Armand S, Aminian K. Locomotion and cadence detection using a single trunk-fixed accelerometer: validity for children with cerebral palsy in daily life-like conditions. Journal of neuroengineering and rehabilitation. 2019;16(1):1–11.
352. Senden R, Grimm B, Heyligers I, Savelberg H, Meijer K. Acceleration-based gait test for healthy subjects: reliability and reference data. Gait & posture. 2009;30(2):192–6.
353. Henriksen M, Lund H, Moe-Nilssen R, Bliddal H, Danneskiod-Samsøe B. Test–retest reliability of trunk accelerometric gait analysis. Gait & posture. 2004;19(3):288–97.
354. Felius RA, Geerars M, Bruijn SM, van Dieën JH, Wouda NC, Punt M. Reliability of IMU-based gait assessment in clinical stroke rehabilitation. Sensors. 2022;22(3):908.
355. Sakhare GM. Determining Predictability of Human Gait Parameters 2017.
356. Quatman-Yates CC, Bonnette S, Hugentobler JA, Médé B, Kiefer AW, Kurowski BG, et al. Postconcussion postural sway variability changes in youth: the benefit of structural variability analyses. Pediatric physical therapy. 2015;27(4):316–27.
357. Purkayastha S, Adair H, Woodruff A, Ryan LJ, Williams B, James E, et al. Balance testing following concussion: postural sway versus complexity index. PM&R. 2019;11(11):1184–92.
358. Troiano M, Thompson X, Boukhechba M, Hertel J, Resch JE. An absence of persistent postural stability deficits following a sport concussion in collegiate athletes. The Journal of Head Trauma Rehabilitation. 2023;38(6):425–33.
359. Schmidt JD, Terry DP, Ko J, Newell KM, Miller LS. Balance regularity among former high school football players with or without a history of concussion. Journal of athletic training. 2018;53(2):109–14.
360. Mooney MG, Cormack S, O'Brien BJ, Morgan WM, McGuigan M. Impact of neuromuscular fatigue on match exercise intensity and performance in elite Australian football. The Journal of Strength & Conditioning Research. 2013;27(1):166–73.
361. Parker TM, Osternig LR, Van Donkelaar P, Chou L. Gait stability following concussion. Medicine and science in sports and exercise. 2006;38(6):1032.
362. Arboix-Alió J, Buscà B, Miró A, Bishop C, Fort-Vanmeerhaeghe A. Ground Reaction Forces, Asymmetries and Performance of Change of Direction Tasks in Youth Elite Female Basketball Players. Sports. 2024;12(1):21.
363. Lee BJ, Blueitt D, Hannon J, Goto S, Garrison C. Movement patterns during a jump-landing task in athletes after sport-related concussion and healthy control individuals. Journal of Athletic Training. 2021;56(12):1306–12.
364. Catena RD, van Donkelaar P, Chou L-S. Different gait tasks distinguish immediate vs. long-term effects of concussion on balance control. Journal of neuroengineering and rehabilitation. 2009;6:25.
365. Parker TM, Osternig LR, van Donkelaar P, Chou L-S. Balance control during gait in athletes and non-athletes following concussion. Medical engineering & physics. 2008;30(8):959–67.
366. Santos-Concejero J, Granados C, Irazusta J, Bidaurrazaga-Letona I, Zabala-Lili J, Tam N, et al. Differences in ground contact time explain the less efficient running economy in North African runners. Biology of sport. 2013;30(3):181–7.
367. Santos-Concejero J, Oliván J, Maté-Muñoz JL, Muniesa C, Montil M, Tucker R, et al. Gait-cycle characteristics and running economy in elite Eritrean and European runners. International Journal of Sports Physiology and Performance. 2015;10(3):381–7.
368. Küderle A, Roth N, Zlatanovic J, Zrenner M, Eskofier B, Kluge F. The placement of foot-mounted IMU sensors does affect the accuracy of spatial parameters during regular walking. Plos one. 2022;17(6):e0269567.

369. Nazarahari M, Khandan A, Khan A, Rouhani H. Foot angular kinematics measured with inertial measurement units: A reliable criterion for real-time gait event detection. Journal of Biomechanics. 2022;130:110880.
370. Soulard J, Vaillant J, Balaguier R, Vuillerme N. Spatio-temporal gait parameters obtained from foot-worn inertial sensors are reliable in healthy adults in single-and dual-task conditions. Scientific Reports. 2021;11(1):10229.
371. Kim BH, Hong SH, Oh IW, Lee YW, Kee IH, Lee SY. Measurement of ankle joint movements using IMUs during running. Sensors. 2021;21(12):4240.
372. Eggers TM, Massard TI, Clothier PJ, Lovell R. Measuring vertical stiffness in sport with accelerometers: exercise caution! The Journal of Strength & Conditioning Research. 2018;32(7):1919–22.
373. Edwards S, White S, Humphreys S, Robergs R, O'Dwyer N. Caution using data from triaxial accelerometers housed in player tracking units during running. Journal of sports sciences. 2019;37(7):810–8.
374. Rowe BH, Yang EH, Gaudet LA, Eliyahu L, Junqueira DR, Beach J, et al. Sex-based differences in outcomes for adult patients presenting to the emergency department with a concussion. Journal of Neurosurgery. 2022;136(1):264–73.
375. Graham R, Rivara FP, Ford MA, Spicer CM, Youth CoS-RCi, Council NR. Concussion recognition, diagnosis, and acute management. Sports-Related Concussions in Youth: Improving the Science, Changing the Culture: National Academies Press (US); 2014.
376. Pierpoint LA, Collins C. Epidemiology of sport-related concussion. Clinics in sports medicine. 2021;40(1):1–18.
377. McNeel C, Clark GM, Davies CB, Major BP, Lum JA. Concussion incidence and time-loss in Australian football: A systematic review. Journal of science and medicine in sport. 2020;23(2):125–33.
378. Echemendia RJ, Brett BL, Broglio S, Davis GA, Giza CC, Guskiewicz KM, et al. Sport concussion assessment tool™ – 6 (SCAT6). British journal of sports medicine. 2023;57(11):622–31.
379. Dunne LA, Cole MH, Cormack SJ, Howell DR, Johnston RD. Validity and Reliability of Methods to Assess Movement Deficiencies Following Concussion: A COSMIN Systematic Review. Sports Medicine-Open. 2023;9(1):76.
380. Davis GA, Makdissi M, Bloomfield P, Clifton P, Cowie C, Echemendia R, et al. Concussion Guidelines in National and International Professional and Elite Sports. Neurosurgery. 2020;87(2):418–25.
381. Laura A M Dunne MHC, Stuart J Cormack, David R Howell, Daniel Chalkley, Ben Horsley, & Rich D Johnston. Reliability of wearable devices for measuring gait parameters on grassed surfaces amongst recreationally active adults. 2024.
382. Association WM. World Medical Association Declaration of Helsinki. Ethical principles for medical research involving human subjects. Bulletin of the world health organization. 2001;79(4):373.
383. Bürkner P-C. brms: An R package for Bayesian multilevel models using Stan. Journal of statistical software. 2017;80:1–28.
384. Lodewyckx T, Kim W, Lee MD, Tuerlinckx F, Kuppens P, Wagenmakers E-J. A tutorial on Bayes factor estimation with the product space method. Journal of Mathematical Psychology. 2011;55(5):331–47.
385. Kass RE, Raftery AE. Bayes factors. Journal of the american statistical association. 1995;90(430):773–95.
386. Catena RD, Van Donkelaar P, Chou L-S. Cognitive task effects on gait stability following concussion. Experimental brain research. 2007;176:23–31.
387. Lynall RC, Mauntel TC, Padua DA, Mihalik JP. Acute lower extremity injury rates increase after concussion in college athletes. Medicine & Science in Sports & Exercise. 2015;47(12):2487–92.

388. Blair S, Lake MJ, Ding R, Sterzing T. Magnitude and variability of gait characteristics when walking on an irregular surface at different speeds. Human movement science. 2018;59:112–20.
389. Martini DN, Sabin MJ, DePesa SA, Leal EW, Negrete TN, Sosnoff JJ, et al. The chronic effects of concussion on gait. Archives of physical medicine and rehabilitation. 2011;92(4):585–9.
390. Beange KH, Chan AD, Beaudette SM, Graham RB. Concurrent validity of a wearable IMU for objective assessments of functional movement quality and control of the lumbar spine. Journal of biomechanics. 2019;97:109356.
391. Celik Y, Stuart S, Woo WL, Godfrey A. Gait analysis in neurological populations: Progression in the use of wearables. Medical Engineering & Physics. 2021;87:9–29.
392. Klöpfer-Krämer I, Brand A, Wackerle H, Müßig J, Kröger I, Augat P. Gait analysis–Available platforms for outcome assessment. Injury. 2020;51:S90–S6.
393. Howell DR, Osternig LR, Chou L-S. Adolescents demonstrate greater gait balance control deficits after concussion than young adults. The American journal of sports medicine. 2015;43(3):625–32.
394. Kim S, Park S. Leg stiffness increases with speed to modulate gait frequency and propulsion energy. Journal of biomechanics. 2011;44(7):1253–8.
395. Walsh GS. Dynamics of modular neuromotor control of walking and running during single and dual task conditions. Neuroscience. 2021;465:1–10.
396. Lehnert M, De Ste Croix M, Zaatar A, Hughes J, Varekova R, Lastovicka O. Muscular and neuromuscular control following soccer-specific exercise in male youth: Changes in injury risk mechanisms. Scandinavian journal of medicine & science in sports. 2017;27(9):975–82.
397. Howell DR, Bonnette S, Diekfuss JA, Grooms DR, Myer GD, Wilson JC, et al. Dual-task gait stability after concussion and subsequent injury: an exploratory investigation. Sensors. 2020;20(21):6297.
398. Coutts AJ, Quinn J, Hocking J, Castagna C, Rampinini E. Match running performance in elite Australian Rules Football. Journal of Science and Medicine in Sport. 2010;13(5):543–8.
399. Deutsch M, Kearney G, Rehrer N. Time–motion analysis of professional rugby union players during match-play. Journal of sports sciences. 2007;25(4):461–72.
400. Fernández-Valdés B, Jones B, Hendricks S, Weaving D, Ramirez-Lopez C, Whitehead S, et al. A novel application of entropy analysis for assessing changes in movement variability during cumulative tackles in young elite rugby league players. Biology of Sport. 2022;40(1):161–70.
401. Howell DR, Seehusen CN, Carry PM, Walker GA, Reinking SE, Wilson JC. An 8-week neuromuscular training program after concussion reduces 1-year subsequent injury risk: a randomized clinical trial. The American journal of sports medicine. 2022;50(4):1120–9.
402. Howell DR, Seehusen CN, Walker GA, Reinking S, Wilson JC. Neuromuscular training after concussion to improve motor and psychosocial outcomes: a feasibility trial. Physical therapy in sport. 2021;52:132–9.

# Appendices

## Appendix 1: Evidence of publication

### Systematic review

Dunne, L.A.M., Cole, M.H., Cormack, S.J. *et al.* Validity and Reliability of Methods to Assess Movement Deficiencies Following Concussion: A COSMIN Systematic Review. *Sports Med - Open* **9**, 76 (2023). https://doi.org/10.1186/s40798-023-00625-0

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## Appendix 2: Post-Concussion Symptom Scale (PCSS)

Player's Name: ________________ Team: ______________ Position: ____________

| SYMPTOM | RATING<br>None Mod. Severe | BASELINE<br>Date: | TESTING 2<br>Date: | TESTING 3<br>Date: | TESTING 4<br>Date: | TESTING 5<br>Date: |
|---|---|---|---|---|---|---|
| Headache | 0 1 2 3 4 5 6 | | | | | |
| Nausea | 0 1 2 3 4 5 6 | | | | | |
| Vomiting | 0 1 2 3 4 5 6 | | | | | |
| Balance problems | 0 1 2 3 4 5 6 | | | | | |
| Dizziness | 0 1 2 3 4 5 6 | | | | | |
| Fatigue | 0 1 2 3 4 5 6 | | | | | |
| Trouble falling asleep | 0 1 2 3 4 5 6 | | | | | |
| Sleeping more than usual | 0 1 2 3 4 5 6 | | | | | |
| Sleeping less than usual | 0 1 2 3 4 5 6 | | | | | |
| Drowsiness | 0 1 2 3 4 5 6 | | | | | |
| Sensitivity to light | 0 1 2 3 4 5 6 | | | | | |
| Sensitivity to noise | 0 1 2 3 4 5 6 | | | | | |
| Irritability | 0 1 2 3 4 5 6 | | | | | |
| Sadness | 0 1 2 3 4 5 6 | | | | | |
| Nervousness | 0 1 2 3 4 5 6 | | | | | |
| Feeling more emotional | 0 1 2 3 4 5 6 | | | | | |
| Numbness or tingling | 0 1 2 3 4 5 6 | | | | | |
| Feeling slowed down | 0 1 2 3 4 5 6 | | | | | |
| Feeling mentally "foggy" | 0 1 2 3 4 5 6 | | | | | |
| Difficulty concentrating | 0 1 2 3 4 5 6 | | | | | |
| Difficulty remembering | 0 1 2 3 4 5 6 | | | | | |
| TOTAL SCORE | | | | | | |

**Appendix 3: Borg 6-20 Rate of perceived exertion (RPE) scale**

| Rating | Descriptor |
|---|---|
| 6 | No exertion at all |
| 7 | Extremely light |
| 8 | |
| 9 | Very light |
| 10 | |
| 11 | Light |
| 12 | |
| 13 | Somewhat hard |
| 14 | |
| 15 | Hard (heavy) |
| 16 | |
| 17 | Very hard |
| 18 | |
| 19 | Extremely hard |
| 20 | Maximal exertion |

## Appendix 4: Ethics approval – Studies 2 & 3

| | |
|---|---|
| Subject | **[2021-214H] - Ethics application approved!** |
| Link to Outlook Item | Click here |
| From | Res Ethics |
| To | Richard Johnston; Laura Dunne |
| Cc | Res Ethics |
| Sent | **12/3/2021 9:06:02 AM** |

Dear Applicant,

Chief Investigator: Mr Richard Johnston
Co-Investigators: A/Prof David Howell, Assoc. Prof. Stuart Cormack, Dr Michael Cole,
Student Researcher: Laura Dunne (Doctoral Student) Admin Contact: Ms Frances Taylor
Ethics Register Number: 2021-214H
Project Title: Using wearable microtechnology devices to monitor movement strategy changes following concussion
Date Approved: 03/12/2021
End Date: 31/10/2023

This is to certify that the above human ethics application has been reviewed by the Australian Catholic University Human Research Ethics Committee (ACU HREC). The application has been approved for the period given above.

Continued approval of this research project is contingent upon the submission of an annual progress report which is due on/before each anniversary of the project approval. A final report is due upon completion of the project. A report proforma can be downloaded from the ACU Research Ethics website.

Researchers are responsible for ensuring that all conditions of approval are adhered to and that any modifications to the protocol, including changes to personnel, are approved prior to implementation. In addition, the ACU HREC must be notified of any reportable matters including, but not limited to, incidents, complaints and unexpected issues.

Researchers are also responsible for ensuring that they adhere to the requirements of the National Statement on Ethical Conduct in Human Research, the Australian Code for the Responsible Conduct of Research and the University's Research Code of Conduct.

Any queries relating to this application should be directed to the Ethics Secretariat (res.ethics@acu.edu.au). Please quote your ethics approval number in all communications with us.

We wish you every success with your research.

Kind regards,

Kylie Pashley
on behalf of ACU HREC Chair, Assoc Prof. Michael Baker

Senior Research Ethics Officer | Research Services | Office of the Deputy Vice-Chancellor (Research)
Australian Catholic University
T: +61 2 9739 2646 E: res.ethics@acu.edu.au

## Appendix 5: Information letter and consent form – Studies 2 & 3

# PARTICIPANT INFORMATION LETTER
# Concussed Participants
# Adult Participants (over 18 years)

**PROJECT TITLE:** Using microtechnology to monitor movement strategy changes following concussion
**APPLICATION NUMBER: (2021- 214H)**
**PRINCIPAL INVESTIGATOR: Dr Rich Johnston**
**STUDENT RESEARCHER: Laura Dunne**
**STUDENT'S DEGREE:** Doctor of Philosophy

Dear Participant,

You are invited to participate in the research project described below.

***What is the project about?***
The research project investigates the ability of commercially worn GPS devices (those worn in playing jerseys) to detect changes in movement strategies following a sports-related concussion. This project aims to assess the validity and reliability of these devices, how long changes in movement last before and following return to play from a sports-related concussion, and finally if these changes are exacerbated when players become fatigued. Results from this project may provide players, coaches, and team health staff with a more objective and clearer way to assess readiness for return to play following a sports-related concussion.

***Who is undertaking the project?***
This project is being conducted by Laura Dunne, and will form the basis for the Doctor of Philosophy degree at Australian Catholic University under the supervision of Dr Rich Johnston. Laura has previously completed a Bachelor of Exercise and Sports Science and a Masters in Clinical Exercise Physiology at Australian Catholic University. She has a strong background in management and exercise prescription for both healthy and clinical populations. Dr Rich Johnston is an applied sport science researcher with a Masters degree in strength and conditioning and a PhD. He has worked as a practitioner for over 10 years with athletes from a number of sports. He has led numerous research projects with athletes, focusing on aspects of injury and performance, with over 50 peer reviewed publications.

***Are there any risks associated with participating in this project?***
Some risks associated with this project may include symptoms re-emerging during the exercise protocol following return from concussion. To mitigate these risks, all participants will have been cleared to return to graduated exercise; all participants will be required to complete a 22-item Post-Concussion Symptom Scale (PCSS) questionnaire prior to data collection periods. If symptoms increase, exercise will be stopped and participants will be referred to a doctor (team or personal doctor) to have their symptoms and recovery re-assessed. The protocol will be conducted in an open grass field away from any potential obstacles that may increase falls risk.

***What will I be asked to do?***

Participants will be asked to complete a questionnaire regarding demographic information including; age, height, weight, sex, physical activity, current sports participation, concussion history, and injury history.

Upon sustaining a medically diagnosed sports-related concussion, and being cleared by a doctor for return to training, participants will complete the following protocol.

Prior to data collection:

- Participants will have a thoracic-mounted GPS device worn in a tight-fitting vest along with additional devices fixed on the skin at the 5th lumbar vertebrae (lower back) and the tibia (ankle) that measure speed and accelerations
- Prior to testing all participants will be required to complete a post-concussion symptom scale (PCSS) questionnaire. This will allow researchers to identify if any participants are still experiencing symptoms of concussion. Those who are still display symptoms will be excluded until asymptomatic
- Participants will then perform a standardised dynamic warm-up including dynamic stretching and 20 to 40-m runs progressively increasing in speed
- The protocol will then be demonstrated to participants, with cones set along the 40-m track indicating acceleration, maintenance, and deceleration periods
- Participants will complete 2 familiarisations of each condition (walk, jog, and run) at a pre-determined set-speed
- During this time, participants will also become familiar with how to use the rating of perceived exertion (RPE) scale

Data collection protocol:

- Participants will be required to perform 3 trials of each condition; walk, jog, and run. These will be performed under self-paced and set-paced conditions
- Rating of perceived exertion (RPE) on a scale of 1-10 will be recorded after each trial from respective participants
- A total of 18 x 40-m trials at varying speeds will be performed during each testing period. These will be performed at the beginning and/or end of training sessions

Participants included in the project will be asked to complete this protocol longitudinally over 1 and/or 2 consecutive competitive seasons. Upon sustaining a medically diagnosed concussion, participants will be asked to complete the protocol upon return to run, return to play, 2 weeks post return to play, and potentially 4 weeks post return to play should significant changes in accelerometer profiles be evident. Those with significant changes will continue to be monitored and assessed at 4-weekly intervals for a subsequent 6-months or until no significant changes in accelerometer profiles are evident (whichever comes first). The protocol will be run at participants training facilities during training times (before and after training) or at Australian Catholic University, Brisbane campus.

This project will involve the recording of individual demographic and accelerometer data. No recordings or images of participants will be required for this project.

***How much time will the project take?***

In total, each testing session will take approximately 20-25 minutes to complete. This will be conducted at the beginning and end of training. Due to the assessment being conducted during warm-up and cool-down periods, other than the fitting of the devices prior to the session (approximately 5-10 minutes), no other time commitments from participants will be required.

This study will run over 1 and/or 2 consecutive seasons. Therefore a time commitment up to approximately 2 years is required for participants.

***What are the benefits of the research project?***
Following concussion, most athletes return to play once asymptomatic and return to normal (pre-concussion) cognitive and balance scores based on normalised clinical assessments. This project may provide a more valid and objective measure of potential movement strategy changes following return to play from a sports-related concussion. Results from this project may better inform players and coaching staff regarding recovery from concussion, and therefore allow for more individualised return to play management.

***Can I withdraw from the study?***
Participation in this study is completely voluntary. You are not under any obligation to participate. If you agree to participate, you can withdraw from the study at any time without adverse consequences. However, withdrawal from the study after analysis of de-identified data (approximately after completing 1 session) may mean participants results cannot be withdrawn from the study analysis.

***Will anyone else know the results of the project?***
Results from this project will be published in academic peer-reviewed journals. Participants will be de-identified in publications of these studies. Data collected throughout the duration of the project will be stored in a password protected electronic database and destroyed after 15 years.

***Will I be able to find out the results of the project?***
A summary of de-identified results will be available for participants to view. This will be provided to coaching and/or school administrative staff with the availability for participants to view and access an electric copy of these.

***Who do I contact if I have questions about the project?***
If you have any questions regarding this project or individual results please contact Laura Dunne
Email: laura.dunne@myacu.edu.au
Mobile: 04 17378563

***What if I have a complaint or any concerns?***
The study has been reviewed by the Human Research Ethics Committee at Australian Catholic University (review number 2021-214H). If you have any complaints or concerns about the conduct of the project, you may write to the Manager of the Human Research Ethics and Integrity Committee care of the Office of the Deputy Vice Chancellor (Research).

Manager, Ethics and Integrity
c/o Office of the Deputy Vice Chancellor (Research)
Australian Catholic University
North Sydney Campus
PO Box 968
NORTH SYDNEY, NSW 2059

Ph.: 02 9739 2519
Fax: 02 9739 2870
Email: resethics.manager@acu.edu.au

Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

***I want to participate! How do I sign up?***

If you wish to participate, please contact Laura Dunne (details provided above) to discuss any potential questions or concerns regarding the study. Once you have discussed this project, please fill out the eligibility criteria form and electronic consent form if eligible.

Yours sincerely,

[redacted]

**Laura Dunne**

# PARTICIPANT INFORMATION LETTER
# Concussed Participants
# Minors (under 18 years)

**PROJECT TITLE:** Using microtechnology to monitor movement strategy changes following concussion
**APPLICATION NUMBER: (2021- 214H)**
**PRINCIPAL INVESTIGATOR: Dr Rich Johnston**
**STUDENT RESEARCHER: Laura Dunne**
**STUDENT'S DEGREE:** Doctor of Philosophy

Dear Participant and parent and/or caregiver,

You are invited to participate in the research project described below.

***What is the project about?***
The research project investigates the ability of commercially worn GPS devices (those worn in playing jerseys) to detect changes in movement strategies following a sports-related concussion. This project aims to assess the validity and reliability of these devices, how long changes in movement last before and following return to play from a sports-related concussion, and finally if these changes are exacerbated when players become fatigued. Results from this project may provide players, coaches, and team health staff with a more objective and clearer way to assess readiness for return to play following a sports-related concussion.

***Who is undertaking the project?***
This project is being conducted by Laura Dunne, and will form the basis for the Doctor of Philosophy degree at Australian Catholic University under the supervision of Dr Rich Johnston. Laura has previously completed a Bachelor of Exercise and Sports Science and a Masters in Clinical Exercise Physiology at Australian Catholic University. She has a strong background in management and exercise prescription for both healthy and clinical populations. Dr Rich Johnston is an applied sport science researcher with a Masters degree in strength and conditioning and a PhD. He has worked as a practitioner for over 10 years with athletes from a number of sports. He has led numerous research projects with athletes, focusing on aspects of injury and performance, with over 50 peer reviewed publications.

***Are there any risks associated with participating in this project?***
Some risks associated with this project may include; symptoms re-emerging during the exercise protocol following return from concussion. To mitigate these risks, all participants will have been cleared to return to graduated exercise; all participants will be required to complete a 22-item Post-Concussion Symptom Scale (PCSS) questionnaire prior to data collection periods. If symptoms increase, exercise will be stopped and participants will be referred to a doctor (team or personal doctor) to have their symptoms and recovery re-assessed. The protocol will be conducted in an open grass field away from any potential obstacles that may increase falls risk.

***What will I be asked to do?***
Participants will be asked to complete a questionnaire regarding demographic information including; age, height, weight, sex, physical activity, current sports participation, concussion history, and injury history.
Upon sustaining a medically diagnosed sports-related concussion, and being cleared by a doctor for return to training, participants will complete the following protocol.

Prior to data collection:

- Participants will have a thoracic-mounted GPS device worn in a tight-fitting vest along with additional devices fixed on the skin at the 5th lumbar vertebrae (lower back) and the tibia (ankle) that measure speed and accelerations
- Prior to testing all participants will be required to complete a post-concussion symptom scale (PCSS) questionnaire. This will allow researchers to identify if any participants are still experiencing symptoms of concussion. Those who are still display symptoms will be excluded until asymptomatic
- Participants will then perform a standardised dynamic warm-up including dynamic stretching and 20 to 40-m runs progressively increasing in speed
- The protocol will then be demonstrated to participants, with cones set along the 40-m track indicating acceleration, maintenance, and deceleration periods
- Participants will complete 2 familiarisations of each condition (walk, jog, and run) at a pre-determined set-speed
- During this time, participants will also become familiar with how to use the rating of perceived exertion (RPE) scale

Data collection protocol:

- Participants will be required to perform 3 trials of each condition; walk, jog, and run. These will be performed under self-paced and set-paced conditions
- Rating of perceived exertion (RPE) on a scale of 1-10 will be recorded after each trial from respective participants
- A total of 18 x 40-m trials at varying speeds will be performed during each testing period. These will be performed at the beginning and/or end of training sessions

Participants included in the project will be asked to complete this protocol longitudinally over 1 and/or 2 consecutive competitive seasons. Upon sustaining a medically diagnosed concussion, participants will be asked to complete the protocol upon return to run, return to play, 2 weeks post return to play, and potentially 4 weeks post return to play should significant changes in accelerometer profiles be evident. Those with significant changes will continue to be monitored and assessed at 4-weekly intervals for a subsequent 6-months or until no significant changes in accelerometer profiles are evident (whichever comes first). The protocol will be run at participants training facilities during training times (before and after training) or at Australian Catholic University, Brisbane campus.

This project will involve the recording of individual demographic and accelerometer data. No recordings or images of participants will be required for this project.

***How much time will the project take?***
In total, each testing session will take approximately 20-25 minutes to complete. This will be conducted at the beginning and end of training. Due to the assessment being conducted during warm-

up and cool-down periods, other than the fitting of the devices prior to the session (approximately 5-10 minutes), no other time commitments from participants will be required.

This study will run over 1 and/or 2 consecutive seasons. Therefore a time commitment up to approximately 2 years is required for participants.

***What are the benefits of the research project?***
Following concussion, most athletes return to play once asymptomatic and return to normal (pre-concussion) cognitive and balance scores based on normalised clinical assessments. This project may provide a more valid and objective measure of potential movement strategy changes following return to play from a sports-related concussion. Results from this project may better inform players and coaching staff regarding recovery from concussion, and therefore allow for more individualised return to play management.

***Can I withdraw from the study?***
Participation in this study is completely voluntary. You are not under any obligation to participate. If you agree to participate, you can withdraw from the study at any time without adverse consequences. However, withdrawal from the study after analysis of de-identified data (approximately after completing 1 session) may mean participants results cannot be withdrawn from the study analysis.

***Will anyone else know the results of the project?***
Results from this project will be published in academic peer-reviewed journals. Participants will be de-identified in publications of these studies. Data collected throughout the duration of the project will be stored in a password protected electronic database and destroyed after 15 years.

***Will I be able to find out the results of the project?***
A summary of de-identified results will be available for participants to view. This will be provided to coaching and/or school administrative staff with the availability for participants to view and access an electric copy of these.

***Who do I contact if I have questions about the project?***
If you have any questions regarding this project or individual results please contact Laura Dunne
Email: laura.dunne@myacu.edu.au
Mobile: 04 17378563

***What if I have a complaint or any concerns?***
The study has been reviewed by the Human Research Ethics Committee at Australian Catholic University (review number 2021-214H). If you have any complaints or concerns about the conduct of the project, you may write to the Manager of the Human Research Ethics and Integrity Committee care of the Office of the Deputy Vice Chancellor (Research).

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Ph.: 02 9739 2519
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Email: resethics.manager@acu.edu.au

Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

***I want to participate! How do I sign up?***

If you wish to participate, please have a parent and/or caregiver contact Laura Dunne (details provided above) to discuss any potential questions or concerns regarding the project. Once you have discussed this project, please have both participant (under 18 years of age) and a parent and/or caregiver complete the eligibility criteria form and the relevant sections of the electronic consent form if eligible.

Yours sincerely,

[redacted]

**Laura Dunne**

# PARTICIPANT INFORMATION LETTER
# Control Participants
# Adult Participants (over 18 years)

**PROJECT TITLE:** Using microtechnology to monitor movement strategy changes following concussion
**APPLICATION NUMBER: (2021- 214H)**
**PRINCIPAL INVESTIGATOR: Dr Rich Johnston**
**STUDENT RESEARCHER: Laura Dunne**
**STUDENT'S DEGREE:** Doctor of Philosophy

Dear Participant,

You are invited to participate in the research project described below.

***What is the project about?***
The research project investigates the ability of commercially worn GPS devices (those worn in playing jerseys) to detect changes in movement strategies following a sports-related concussion. This project aims to assess the validity and reliability of these devices, how long changes in movement last before and following return to play from a sports-related concussion, and finally if these changes are exacerbated when players become fatigued. Results from this project may provide players, coaches, and team health staff with a more objective and clearer way to assess readiness for return to play following a sports-related concussion.

***Who is undertaking the project?***
This project is being conducted by Laura Dunne, and will form the basis for the Doctor of Philosophy degree at Australian Catholic University under the supervision of Dr Rich Johnston. Laura has previously completed a Bachelor of Exercise and Sports Science and a Masters in Clinical Exercise Physiology at Australian Catholic University. She has a strong background in management and exercise prescription for both healthy and clinical populations. Dr Rich Johnston is an applied sport science researcher with a Masters degree in strength and conditioning and a PhD. He has worked as a practitioner for over 10 years with athletes from a number of sports. He has led numerous research projects with athletes, focusing on aspects of injury and performance, with over 50 peer reviewed publications.

***Are there any risks associated with participating in this project?***
Some risks associated with this project may include those associated with running, including musculoskeletal strain or injury. To mitigate this, all participants will perform a standardised warm-up prior to data collection. Participants who sustained a lower limb injury within 3 months of data collection will not be eligible to participate. To assess any non-concussive impacts participants will also be required to complete a 22-item Post-Concussion Symptom Scale (PCSS) questionnaire prior to data collection periods. If symptoms are present, exercise will be stopped, if symptoms persist, participants will be referred to a doctor (team or personal doctor) to have their symptoms assessed. The protocol will be conducted in an open grass field away from any potential obstacles that may increase falls risk.

***What will I be asked to do?***

Participants will be asked to complete a questionnaire regarding demographic information including; age, height, weight, sex, physical activity, current sports participation, concussion history, and injury history.

Participants in the control group will be required to complete the protocol when an age, sex, and sport matched athlete sustains a concussion. Participants will complete the following protocol.

Prior to data collection:

- Participants will have a thoracic-mounted GPS device worn in a tight-fitting vest along with additional devices fixed on the skin at the 5th lumbar vertebrae (lower back) and the tibia (ankle) that measure speed and accelerations
- Prior to testing all participants will be required to complete a post-concussion symptom scale (PCSS) questionnaire. This will allow researchers to identify if any participants are experiencing symptoms consistent with concussion. Those who display symptoms will be excluded until asymptomatic
- Participants will then perform a standardised dynamic warm-up including dynamic stretching and 20 to 40-m runs progressively increasing in speed
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- During this time, participants will also become familiar with how to use the rating of perceived exertion (RPE) scale

Data collection protocol:

- Participants will be required to perform 3 trials of each condition; walk, jog, and run. These will be performed under self-paced and set-paced conditions
- Rating of perceived exertion (RPE) on a scale of 1-10 will be recorded after each trial from respective participants
- A total of 18 x 40-m trials at varying speeds will be performed during each testing period. These will be performed at the beginning and/or end of training sessions

Participants included in the project will be asked to complete this protocol longitudinally over 1 and/or 2 consecutive competitive seasons. Participants will complete the protocol at the same time points as their age, and sex matched concussed partner. This included testing at the following times; return to run, return to play, 2 weeks post return to play, and potentially 4 weeks post return to play should significant changes in accelerometer profiles be evident in the concussed partner. Control participants will also conduct additional testing sessions at 4-weekly intervals for 6-months. This is to allow researchers to measure the consistency of accelerometer profiles in non-concussed athletes. The protocol will be run at participants training facilities during training times (before and/or after training) or at Australian Catholic University, Brisbane campus.

This project will involve the recording of individual demographic and accelerometer data. No recordings or images of participants will be required for this project.

***How much time will the project take?***
In total, each testing session will take approximately 20-25 minutes to complete. This will be conducted at the beginning and end of training. Due to the assessment being conducted during warm-up and cool-down periods, other than the fitting of the devices prior to the session (approximately 5-10 minutes), no other time commitments from participants will be required.

This study will run over 1 and/or 2 consecutive seasons. Therefore a time commitment up to approximately 2 years is required for participants.

***What are the benefits of the research project?***
Following concussion, most athletes return to play once asymptomatic and return to normal (pre-concussion) cognitive and balance scores based on normalised clinical assessments. This project may provide a more valid and objective measure of potential movement strategy changes following return to play from a sports-related concussion. Results from this project may better inform players and coaching staff regarding recovery from concussion, and therefore allow for more individualised return to play management.

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***Will anyone else know the results of the project?***
Results from this project will be published in academic peer-reviewed journals. Participants will be de-identified in publications of these studies. Data collected throughout the duration of the project will be stored in a password protected electronic database and destroyed after 15 years.

***Will I be able to find out the results of the project?***
A summary of de-identified results will be available for participants to view. This will be provided to coaching and/or school administrative staff with the availability for participants to view and access an electric copy of these.

***Who do I contact if I have questions about the project?***
If you have any questions regarding this project or individual results please contact Laura Dunne

Email: laura.dunne@myacu.edu.au
Mobile: 04 17378563

***What if I have a complaint or any concerns?***
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Manager, Ethics and Integrity
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Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

***I want to participate! How do I sign up?***

If you wish to participate, please contact Laura Dunne (details provided above) to discuss any potential questions or concerns regarding the study. Once you have discussed this project, please fill out the eligibility criteria form and electronic consent form if eligible.

Yours sincerely,

[redacted]

**Laura Dunne**

# PARTICIPANT INFORMATION LETTER
# Control Participants
# Minors (under 18 years)

**PROJECT TITLE:** Using microtechnology to monitor movement strategy changes following concussion
**APPLICATION NUMBER: (2021- 214H)**
**PRINCIPAL INVESTIGATOR: Dr Rich Johnston**
**STUDENT RESEARCHER: Laura Dunne**
**STUDENT'S DEGREE:** Doctor of Philosophy

Dear Participant and parent and/or caregiver,

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***What is the project about?***
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Data collection protocol:

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***How much time will the project take?***
In total, each testing session will take approximately 20-25 minutes to complete. This will be conducted at the beginning and end of training. Due to the assessment being conducted during warm-up and cool-down periods, other than the fitting of the devices prior to the session (approximately 5-10 minutes), no other time commitments from participants will be required.

This study will run over 1 and/or 2 consecutive seasons. Therefore a time commitment up to approximately 2 years is required for participants.

***What are the benefits of the research project?***
Following concussion, most athletes return to play once asymptomatic and return to normal (pre-concussion) cognitive and balance scores based on normalised clinical assessments. This project may provide a more valid and objective measure of potential movement strategy changes following return to play from a sports-related concussion. Results from this project may better inform players and coaching staff regarding recovery from concussion, and therefore allow for more individualised return to play management.

***Can I withdraw from the study?***
Participation in this study is completely voluntary. You are not under any obligation to participate. If you agree to participate, you can withdraw from the study at any time without adverse consequences. However, withdrawal from the study after analysis of de-identified data (approximately after completing 1 session) may mean participants results cannot be withdrawn from the study analysis.

***Will anyone else know the results of the project?***
Results from this project will be published in academic peer-reviewed journals. Participants will be de-identified in publications of these studies. Data collected throughout the duration of the project will be stored in a password protected electronic database and destroyed after 15 years.

***Will I be able to find out the results of the project?***
A summary of de-identified results will be available for participants to view. This will be provided to coaching and/or school administrative staff with the availability for participants to view and access an electric copy of these.

***Who do I contact if I have questions about the project?***
If you have any questions regarding this project or individual results please contact Laura Dunne

Email: laura.dunne@myacu.edu.au
Mobile: 04 17378563

***What if I have a complaint or any concerns?***
The study has been reviewed by the Human Research Ethics Committee at Australian Catholic University (review number 2021-214H). If you have any complaints or concerns about the conduct of the project, you may write to the Manager of the Human Research Ethics and Integrity Committee care of the Office of the Deputy Vice Chancellor (Research).

Manager, Ethics and Integrity
c/o Office of the Deputy Vice Chancellor (Research)

Australian Catholic University
North Sydney Campus
PO Box 968
NORTH SYDNEY, NSW 2059
Ph.: 02 9739 2519
Fax: 02 9739 2870
Email: resethics.manager@acu.edu.au

Any complaint or concern will be treated in confidence and fully investigated. You will be informed of the outcome.

***I want to participate! How do I sign up?***

If you wish to participate, please have a parent and/or caregiver contact Laura Dunne (details provided above) to discuss any potential questions or concerns regarding the project. Once you have discussed this project, please have both participant (under 18 years of age) and a parent and/or caregiver complete the eligibility criteria form and relevant sections of the electronic consent form if eligible.

Yours sincerely,

[redacted]

**Laura Dunne**

## Consent from – Studies 2 & 3

CONSENT FORM
Concussion Group
Adult Participants (Over 18 years of age)
*Copy for Researcher / Copy for Participant to Keep*

TITLE OF PROJECT: Using microtechnology to monitor movement strategy changes following concussion

APPLICATION NUMBER: (2021-214H)

PRINCIPAL INVESTIGATOR: Dr Rich Johnston (supervisor)

STUDENT RESEARCHER: Laura Dunne

I *................................................. (the participant)* have read (*or, where appropriate, have had read to me)* and understood the information provided in the Letter to Participants. Any questions I have asked have been answered to my satisfaction.

I agree to participate in this research project and the collection of demographic data, and understand the requirements of the protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand my participation in this project will only be allowed once I have been medically cleared to return to activities following a sports-related concussion.

I understand I will be required to complete the protocol at the following time points; return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play. I understand that if significant changes in accelerometer profiles are evident, then I will complete the protocol at 4-weeklly intervals for a subsequent 6-months or until no significant changes in accelerometer profiles are evident (whichever comes first). I understand this protocol will be conducted during training times (before and/or after) at my training facility or at Australian Catholic University, Brisbane facilities, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes. I understand the longitudinal time commitment of this project may be up to 2 years.

I realise that I can withdraw my consent at any time (without adverse consequences), however understand that my results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify me in any way.

NAME OF PARTICIPANT: ..........................................................................................................

SIGNATURE: ........................................................................................... DATE:

................................

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): ................................................................

DATE:

………………………………

(and, if applicable)

SIGNATURE OF STUDENT RESEARCHER: ...............................................................................................

DATE:

……………………………

---

PARENT/GUARDIAN CONSENT FORM
Concussion Group
Minors (Under 18 years of age)
*Copy for Researcher / Copy for Participant to Keep*

TITLE OF PROJECT: Using microtechnology to monitor movement strategy changes following concussion

APPLICATION NUMBER: (2021-214H)

PRINCIPAL INVESTIGATOR: Dr Rich Johnston (supervisor)

STUDENT RESEARCHER: Laura Dunne

I .................................................. (*the parent/guardian*) have read (*or, where appropriate*, have had read to me) and understood the information provided in the Letter to the Participants. Any questions I have asked have been answered to my satisfaction.

I agree that my child, nominated below, may participate in this activity with the requirements of the protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand my child's participation in this project will only be allowed once they have been medically cleared to return to activities following a sports-related concussion.

I understand they will be required to complete the protocol at the following time points; return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play. I understand that if significant changes in accelerometer profiles are evident, then my child will complete the protocol at 4-weeklly intervals for a subsequent 6-months or until no significant changes in accelerometer profiles are evident (whichever comes first).I understand this protocol will be conducted during training times (before and/or after) at their training facility or at Australian Catholic University, Brisbane facilities, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes. I understand the longitudinal time commitment of this project may be up to 2 years.

I realise that I can withdraw my consent at any time (without adverse consequences), however understand that my child's results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify my child in any way.

NAME OF PARENT/GUARDIAN: ....................................................................................................

SIGNATURE: ..................................................................................... DATE:

NAME OF CHILD:
………………………………………………………………………………………………………………………………………..

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): ……………………………. DATE: …………………………….

SIGNATURE OF STUDENT RESEARCHER (if applicable): ……………………………………….. DATE: ……………..……………

**ASSENT OF PARTICIPANTS AGED UNDER 18 YEARS**

I ……………………… (*the participant aged under 18 years*) understand what this research project is designed to explore. What I will be asked to do has been explained to me.

I agree to take part in the collection of demographic data and the following protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand my participation in this project will only be allowed once I have been medically cleared to return to activities following a sports-related concussion.

I understand I will be required to complete the protocol at the following time points; return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play. I understand that if significant changes in accelerometer profiles are evident, then I will complete the protocol at 4-weeklly intervals for a subsequent 6-months or until no significant changes in accelerometer profiles are evident (whichever comes first).

I understand this protocol will be conducted during training times and at my training facility, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes. I realise that I can withdraw at any time without having to give a reason for my decision. However understand that my results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify my child in any way.

NAME OF PARTICIPANT AGED UNDER 18: ..............................................................................................

SIGNATURE: ……………………………………………………………………………………………………… DATE: ………………………………..

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): .................................................................
DATE:

SIGNATURE OF STUDENT RESEARCHER (if applicable)]: ..........................................................................

DATE:

CONSENT FORM
Control Group
Adult Participants (Over 18 years of age)
*Copy for Researcher / Copy for Participant to Keep*

TITLE OF PROJECT: Using microtechnology to monitor movement strategy changes following concussion

APPLICATION NUMBER: (2021-214H)

PRINCIPAL INVESTIGATOR (or SUPERVISOR): Dr Rich Johnston (supervisor)

STUDENT RESEARCHER: Laura Dunne

I ................................................. *(the participant)* have read (*or, where appropriate, have had read to me)* and understood the information provided in the Letter to Participants. Any questions I have asked have been answered to my satisfaction.

I agree to participate in this research project and the collection of demographic data, and understand the requirements of the protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand the protocol will be conducted at set time points relative to the participant from the concussed group. Specifically, return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play. I also understand I will complete additional testing sessions at 4-weekly intervals for and additional 6-months.

I understand this protocol will be conducted during training times (before and/or after) at my training facility or at Australian Catholic University, Brisbane facilities, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes. I understand the longitudinal time commitment of this project may be up to 2 years.

I realise that I can withdraw my consent at any time (without adverse consequences), however understand that my results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify me in any way.

NAME OF PARTICIPANT: ..............................................................................................................

SIGNATURE: ............................................................................................ DATE:

................................

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): ..................................................................

DATE:

…………………………….

(and, if applicable)

SIGNATURE OF STUDENT RESEARCHER: ........................................................................................

DATE:

…………………………..

PARENT/GUARDIAN CONSENT FORM
Control Group
Minors (Under 18 years of age)
*Copy for Researcher / Copy for Participant to Keep*

TITLE OF PROJECT: Using microtechnology to monitor movement strategy changes following concussion

APPLICATION NUMBER: (2021-214H)

PRINCIPAL INVESTIGATOR (or SUPERVISOR): Dr Rich Johnston (supervisor)

STUDENT RESEARCHER: Laura Dunne

I .................................................. (*the parent/guardian*) have read (*or, where appropriate*, have had read to me) and understood the information provided in the Letter to the Participants. Any questions I have asked have been answered to my satisfaction.

I agree that my child, nominated below, may participate in this activity with the requirements of the protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand my child will be required to complete the protocol at the following time points; return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play, and these will be relative to the concussed participant they are paired with. I also understand my child will be required to conduct additional testing sessions at 4-weekly intervals for an additional 6-months. I understand this protocol will be conducted during training times (before and/or after) and at their training facility or Australian Catholic University, Brisbane facilities, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes. I understand the longitudinal time commitment of this project may be up to 2 years.

I realise that I can withdraw my consent at any time (without adverse consequences), however understand that my child's results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify my child in any way.

NAME OF PARENT/GUARDIAN: ..........................................................................................................

SIGNATURE: .............................................................................… DATE:

NAME OF CHILD:
……………………………………………………………………………………………………………………………………..

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): ……………………………. DATE: …………………………….

SIGNATURE OF STUDENT RESEARCHER (if applicable): ………………………………………….. DATE: ……………………………

**ASSENT OF PARTICIPANTS AGED UNDER 18 YEARS**

I …………………… (*the participant aged under 18 years*) understand what this research project is designed to explore. What I will be asked to do has been explained to me.

I agree to take part in the collection of demographic data and the following protocol;

- Assessment of symptom severity following clearance to return from concussion
- Wearing of thoracic, lumbar, and tibia-mounted wearable microtechnology devices
- Completing 3 x walk, jog, and run protocols under both set- and self-paced speed conditions over a 40-m distance
- Recording of my rating of perceived exertion (RPE) following each trial

I understand I will be required to complete the protocol at the following time points; return to run, return to play, 2 weeks post return to play, and 4 weeks post return to play, and these will be relative to the concussed participant I am paired with. I also understand I will be required to do additional testing sessions at 4-weekly intervals for an additional 6-months.

I understand this protocol will be conducted during training times and at my training facility or at Australian Catholic University, Brisbane facilities, with an estimated time commitment for fixing of devices and completing the protocol to be approximately 20-25 minutes.

I realise that I can withdraw at any time without having to give a reason for my decision. However understand that my results (de-identified) may still be used in the project analysis. I agree that research data collected for the study may be published or may be provided to other researchers or coaching/medical staff in a form that does not identify my child in any way.

NAME OF PARTICIPANT AGED UNDER 18: ..................................................................................

SIGNATURE: ……………………………………………………………………………………………………… DATE: ………………………………..

SIGNATURE OF PRINCIPAL INVESTIGATOR (or SUPERVISOR): ..................................................................
DATE:

SIGNATURE OF STUDENT RESEARCHER (if applicable)]: ...........................................................................

DATE:

**Appendix 6: R Studio code used for analysis** 2024-10-15

The below code highlights the Bayesian mixed effects models for the analysis between the control and concussed participants

```
######run bayesian models on the data-------------------------------------------
####RPE data
RPE_brms_data = RPEData %>%
  group_by(ID, Assessment, Group)%>%
  summarise(RPE = round(mean(RPE),0))%>%
  mutate(Assessment = as.factor(Assessment),
         Group = factor(Group, levels = c("Control", "Concussed")))

RPE_brms = brm(RPE ~ Group * Assessment + (1|ID),
               chains = 4, iter = 3000,
               control = list(max_treedepth = 15, adapt_delta = 0.9),
               family = sratio(),
               data = RPE_brms_data)
summary(RPE_brms)

#run a model with no priors
Ent_brms_prior_np = brm(EntVert ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
               chains = 2, iter = 4000,
               control = list(max_treedepth = 15),
               data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val))

#model summary
summary(Ent_brms_prior_np)

#Define weakly informative priors for the vertical entropy model
priors = c(
  set_prior("", class = "Intercept"),
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed"),
  set_prior("normal(-0.05, 0.02)", class = "b", coef = "ActivityJog"),
  set_prior("normal(0.10, 0.03)", class = "b", coef = "ActivityRun"),
  set_prior("", class = "b", coef = "Days_Since_Conc"),  # Flat prior for Days_Since_Conc
  set_prior("normal(0.1, 0.05)", class = "b", coef = "GroupConcussed:ActivityJog"),
  set_prior("normal(0.2, 0.05)", class = "b", coef = "GroupConcussed:ActivityRun"),
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"),  # Flat prior for Gr
oupConcussed:Days_Since_Conc
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"),  # Flat prior for Activ
ityJog:Days_Since_Conc
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"),  # Flat prior for Activ
ityRun:Days_Since_Conc
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"),  # Flat
prior for GroupConcussed:ActivityJog:Days_Since_Conc
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc")  # Flat p
rior for GroupConcussed:ActivityRun:Days_Since_Conc
)

# Fit the model with the specified priors
Ent_prior_brms = brm(
  formula = EntVert ~ Group * Activity * Days_Since_Conc + (1 | PlayerID),
  data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val),
  chains = 4,
  iter = 4000,
  control = list(max_treedepth = 15),
  prior = priors
)


prior_summary(Ent_prior_brms)
```

```r
# Print the summary of the model
summary(Ent_prior_brms)
pp_check(Ent_prior_brms)

#Lumbar model
Ent_prior_brms_lum = brm(
  formula = EntVert ~ Group * Activity * Days_Since_Conc + (1 | PlayerID),
  data = dplyr::filter(Data_Lumbar, PlayerID %in% Players_for_Val),
  chains = 4,
  iter = 4000,
  control = list(max_treedepth = 15),
  prior = priors
)

pp_check(Ent_prior_brms_lum, ndraws = 1e2)

#create a tidy output of the data
Ent_prior_brms_output = bind_rows(summary(Ent_prior_brms)$fixed, summary(Ent_prior_brms_lum
)$fixed) %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
    rename(`Bulk ESS` = Bulk_ESS,
           `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Sample entropy, vertical acceleration', .before = 1)

#run a hypothesis test for a difference between groups at baseline
EntVert_hypotheses = bind_rows(
hypothesis(Ent_prior_brms, c("GroupConcussed > 0"))$hypothesis,
hypothesis(Ent_prior_brms, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
hypothesis(Ent_prior_brms, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
hypothesis(Ent_prior_brms, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
hypothesis(Ent_prior_brms, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcussed:D
ays_Since_Conc < 0"))$hypothesis,
hypothesis(Ent_prior_brms, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcussed:D
ays_Since_Conc < 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed > 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcuss
ed:Days_Since_Conc < 0"))$hypothesis,
hypothesis(Ent_prior_brms_lum, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcuss
ed:Days_Since_Conc < 0"))$hypothesis
)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

####mediolateral entropy
EML_priors = c(
  set_prior("normal(0.5, 0.05)", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed"), # Prior for GroupC
oncussed effect
  set_prior("normal(0.2, 0.1)", class = "b", coef = "ActivityJog"), # Prior for ActivityJog
effect
```

```
  set_prior("normal(0.3, 0.1)", class = "b", coef = "ActivityRun"), # Prior for ActivityRun
effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(0.1, 0.1)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prior
for GroupConcussed:ActivityJog interaction
  set_prior("normal(0.15, 0.1)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior
for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
  )

#run the model at thoracic
EntML_brms = brm(EntML ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                 chains = 4, iter = 4000,
                 control = list(max_treedepth = 15),
               data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val),
               prior = EML_priors)

summary(EntML_brms)
prior_summary(EntML_brms)
pp_check(EntML_brms, draw_ids = 'GroupConcussed')

#lumbar
EntML_brms_lum = brm(EntML ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                 chains = 4, iter = 4000,
                 control = list(max_treedepth = 15),
                 data = dplyr::filter(Data_Lumbar, PlayerID %in% Players_for_Val),
                 prior = EML_priors)

summary(EntML_brms_lum)
prior_summary(EntML_brms_lum)
pp_check(EntML_brms_lum, type = "dens_overlay")

#mdeiolateral entropy
EntML_brms_output = bind_rows(summary(EntML_brms)$fixed, summary(EntML_brms_lum)$fixed) %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Sample entropy, mediolateral acceleration', .before = 1)


#Hypothesis tests for mediolateral entropy
EntML_hypotheses = bind_rows(
  hypothesis(EntML_brms, c("GroupConcussed > 0"))$hypothesis,
  hypothesis(EntML_brms, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
  hypothesis(EntML_brms, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
  hypothesis(EntML_brms, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
  hypothesis(EntML_brms, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcussed:Day
```

```
s_Since_Conc < 0"))$hypothesis,
  hypothesis(EntML_brms, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcussed:Day
s_Since_Conc < 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed > 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcussed
:Days_Since_Conc < 0"))$hypothesis,
  hypothesis(EntML_brms_lum, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcussed
:Days_Since_Conc < 0"))$hypothesis
)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))



####steptime
#these priors were set based on large reductions in gait velocity chrome-extension://efaidn
bmnnnibpcajpcglclefindmkaj/https://www.jocn-journal.com/action/showPdf?pii=S0967-5868%2822%
2900457-X
#small differences were set for walk, jog and run at baseline
Steptime_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed"), # Prior for GroupC
oncussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("normal(0, 0.05)", class = "b", coef = "Days_Since_Conc"), # Flat prior for Day
s_Since_Conc effect
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prio
r for GroupConcussed:ActivityJog interaction
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prio
r for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#steptime at thoracic
Steptime_brms = brm(Steptime_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                    chains = 4, iter = 4000,
                    control = list(max_treedepth = 15),
                 data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val),
                 prior = Steptime_priors)
summary(Steptime_brms)

#lumbar spine
Steptime_brms_lum = brm(Steptime_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                        chains = 4, iter = 4000,
                        control = list(max_treedepth = 15),
                        data = dplyr::filter(Data_LumbarStep, PlayerID %in% Players_for_Val),
                        prior = Steptime_priors)
summary(Steptime_brms_lum)
```

```
#create the output from both models
Steptime_brms_output = bind_rows(summary(Steptime_brms)$fixed, summary(Steptime_brms_lum)$f
ixed) %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Average time between steps (s)', .before = 1)

#hypothesis test for step time
Steptime_hypothesis = bind_rows(hypothesis(Steptime_brms, c("GroupConcussed > 0"))$hypothes
is,
          hypothesis(Steptime_brms, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
          hypothesis(Steptime_brms, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
          hypothesis(Steptime_brms, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
          hypothesis(Steptime_brms, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupCo
ncussed:Days_Since_Conc < 0"))$hypothesis,
          hypothesis(Steptime_brms, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupCo
ncussed:Days_Since_Conc < 0"))$hypothesis,
          hypothesis(Steptime_brms_lum, c("GroupConcussed > 0"))$hypothesis,
          hypothesis(Steptime_brms_lum, c("GroupConcussed:ActivityJog > 0"))$hypothesis,
          hypothesis(Steptime_brms_lum, c("GroupConcussed:ActivityRun > 0"))$hypothesis,
          hypothesis(Steptime_brms_lum, c("GroupConcussed:Days_Since_Conc < 0"))$hypothesis
,
          hypothesis(Steptime_brms_lum, c("GroupConcussed:ActivityJog:Days_Since_Conc + Gro
upConcussed:Days_Since_Conc < 0"))$hypothesis,
          hypothesis(Steptime_brms_lum, c("GroupConcussed:ActivityRun:Days_Since_Conc + Gro
upConcussed:Days_Since_Conc < 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

#steplength
#set the priors for steplength
#based on this study https://www.sciencedirect.com/science/article/pii/S0003999320302185?vi
a%3Dihub#sec2 and this one https://www.sciencedirect.com/science/article/pii/S1440244017309
921?via%3Dihub#sec0015
#shorter step length during walking with a large (d = 1) and moderate (d = 0.7) effect size
difference
Steplength_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(-0.2, 0.1)", class = "b", coef = "GroupConcussed"), # Prior for GroupCo
ncussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(-0.2, 0.1)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prior
for GroupConcussed:ActivityJog interaction
  set_prior("normal(-0.2, 0.1)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior
for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
```

```r
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#run steplength model thoracic
Steplength_brms = brm(Step_length_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                      chains = 4, iter = 4000,
                      control = list(max_treedepth = 15),
                     data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val),
                     prior = Steplength_priors)

summary(Steplength_brms)


#compute the output
Steplength_brms_output = summary(Steplength_brms)$fixed %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ))%
>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Tail_ESS)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Step length (m)', .before = 1)

#hypothesis test for step length
Steplength_hypothesis = bind_rows(hypothesis(Steplength_brms, c("GroupConcussed < 0"))$hypo
thesis,
                                hypothesis(Steplength_brms, c("GroupConcussed:ActivityJog <
0"))$hypothesis,
                                hypothesis(Steplength_brms, c("GroupConcussed:ActivityRun <
0"))$hypothesis,
                                hypothesis(Steplength_brms, c("GroupConcussed:Days_Since_Co
nc > 0"))$hypothesis,
                                hypothesis(Steplength_brms, c("GroupConcussed:ActivityJog:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                                hypothesis(Steplength_brms, c("GroupConcussed:ActivityRun:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

####cadence
#set the priors for cadence
#based on this study https://www.sciencedirect.com/science/article/pii/S1440244017309921?vi
a%3Dihub#sec0015
#shorter step length during walking with a large d = 2 effect size difference, applied simi
lar to prior to jog and run
Cadence_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(-0.1, 0.03)", class = "b", coef = "GroupConcussed"), # Prior for GroupC
oncussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(-0.11, 0.03)", class = "b", coef = "GroupConcussed:ActivityJog"), # Pri
or for GroupConcussed:ActivityJog interaction
  set_prior("normal(-0.11, 0.03)", class = "b", coef = "GroupConcussed:ActivityRun"), # Pri
```

```
or for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#run cadence model for thoracic spine
Cadence_brms = brm(Cadence ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                   chains = 4, iter = 4000,
                   control = list(max_treedepth = 15),
                   data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val),
                   prior = Cadence_priors)

summary(Cadence_brms)
pp_check(Cadence_brms)

#for lumbar spine
Cadence_brms_lum = brm(Cadence ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                   chains = 4, iter = 4000,
                   control = list(max_treedepth = 15),
                   data = dplyr::filter(Data_LumbarStep, PlayerID %in% Players_for_Val),
                   prior = Cadence_priors)

summary(Cadence_brms_lum)

#output for cadence
Cadence_brms_output = bind_rows(summary(Cadence_brms)$fixed, summary(Cadence_brms_lum)$fixe
d) %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Cadence (steps.s.-1)', .before = 1)

#hypothesis test for cadence
Cadence_hypothesis = bind_rows(hypothesis(Cadence_brms, c("GroupConcussed < 0"))$hypothesis
,
                               hypothesis(Cadence_brms, c("GroupConcussed:ActivityJog < 0")
)$hypothesis,
                               hypothesis(Cadence_brms, c("GroupConcussed:ActivityRun < 0")
)$hypothesis,
                               hypothesis(Cadence_brms, c("GroupConcussed:Days_Since_Conc >
0"))$hypothesis,
                               hypothesis(Cadence_brms, c("GroupConcussed:ActivityJog:Days_
Since_Conc + GroupConcussed:Days_Since_Conc> 0"))$hypothesis,
                               hypothesis(Cadence_brms, c("GroupConcussed:ActivityRun:Days_
Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(Cadence_brms_lum, c("GroupConcussed < 0"))$hypoth
esis,
                               hypothesis(Cadence_brms_lum, c("GroupConcussed:ActivityJog <
0"))$hypothesis,
```

```
                                hypothesis(Cadence_brms_lum, c("GroupConcussed:ActivityRun <
0"))$hypothesis,
                                hypothesis(Cadence_brms_lum, c("GroupConcussed:Days_Since_Co
nc > 0"))$hypothesis,
                                hypothesis(Cadence_brms_lum, c("GroupConcussed:ActivityJog:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                                hypothesis(Cadence_brms_lum, c("GroupConcussed:ActivityRun:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

save.image()

#vertical accerlation at each step
#priors for vertical acceleration
AcclVert_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(-0.2, 0.05)", class = "b", coef = "GroupConcussed"), # Prior for GroupC
oncussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("normal(0, 0.02)", class = "b", coef = "Days_Since_Conc"), # Flat prior for Day
s_Since_Conc effect
  set_prior("normal(-0.3, 0.05)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prio
r for GroupConcussed:ActivityJog interaction
  set_prior("normal(-0.3, 0.05)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prio
r for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

AcclVert_priors2 = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(-0.3, 0.1)", class = "b", coef = "GroupConcussed"), # Prior for GroupCo
ncussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("normal(0, 0.1)", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days
_Since_Conc effect
  set_prior("normal(-0.3, 0.2)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prior
for GroupConcussed:ActivityJog interaction
  set_prior("normal(-0.3, 0.2)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior
for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for Activi
tyJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat p
rior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)
```

```
####vertical acceleration
DataGPSStep %>%
  group_by(Group, Activity)%>%
  dplyr::filter(Assessment == 1  & PlayerID %in% Players_for_Val)%>%
  summarise(mean = mean(Accl_vert_av, na.rm = T),
            sd = sd(Accl_vert_av, na.rm = T))%>%
  pivot_wider(values_from = c(mean:sd), names_from = Group)%>%
  mutate(diff = mean_Concussed-mean_Control,
         ES = (mean_Concussed-mean_Control)/(sd_Control+sd_Concussed/2))

#thoracic spine
AcclVert_brms_np = brm(Accl_vert_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                     chains = 4, iter = 4000,
                     control = list(max_treedepth = 15),
                     data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val))

summary(AcclVert_brms_np)

#thoracic spine
AcclVert_brms2 = brm(Accl_vert_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                    chains = 4, iter = 4000,
                    control = list(max_treedepth = 15),
                    data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val),
                    prior = AcclVert_priors2)

summary(AcclVert_brms2)

#lumbar spine
AcclVert_brms_lum2 = brm(Accl_vert_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                    chains = 4, iter = 4000,
                    control = list(max_treedepth = 15),
                    data = dplyr::filter(Data_LumbarStep, PlayerID %in% Players_for_Val),
                    prior = AcclVert_priors2)

summary(AcclVert_brms_lum2)
pp_check(AcclVert_brms2)

#lumbar spine with no priors
AcclVert_brms_lum_np = brm(Accl_vert_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID)
,
                        chains = 4, iter = 4000,
                        control = list(max_treedepth = 15),
                        data = dplyr::filter(Data_LumbarStep, PlayerID %in% Players_for_Va
l))
summary(AcclVert_brms_lum_np)

#output for the models
AcclVert_brms_output = bind_rows(summary(AcclVert_brms2)$fixed, summary(AcclVert_brms_lum2)
$fixed) %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Vertical acceleration (g)', .before = 1)
```

```r
#hypothesis test for vertical acceleraiton
AcclVert_hypothesis = bind_rows(hypothesis(AcclVert_brms2, c("GroupConcussed < 0"))$hypothesis,
                               hypothesis(AcclVert_brms2, c("GroupConcussed:ActivityJog < 0"))$hypothesis,
                               hypothesis(AcclVert_brms2, c("GroupConcussed:ActivityRun < 0"))$hypothesis,
                               hypothesis(AcclVert_brms2, c("GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(AcclVert_brms2, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(AcclVert_brms2, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed < 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed:ActivityJog < 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed:ActivityRun < 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed:ActivityJog:Days_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis,
                               hypothesis(AcclVert_brms_lum2, c("GroupConcussed:ActivityRun:Days_Since_Conc + GroupConcussed:Days_Since_Conc > 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))


#priors for mediolateral acceleration
AcclML_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0.05, 0.02)", class = "b", coef = "GroupConcussed"), # Prior for GroupConcussed effect
  set_prior("", class = "b", coef = "ActivityJog"), # Prior for ActivityJog effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc effect
  set_prior("normal(0.1, 0.02)", class = "b", coef = "GroupConcussed:ActivityJog"), # Prior for GroupConcussed:ActivityJog interaction
  set_prior("normal(0.1, 0.02)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for GroupConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityJog:Days_Since_Conc"), # Flat prior for ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for ActivityRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityJog:Days_Since_Conc"), # Flat prior for GroupConcussed:ActivityJog:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat prior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

####mediolateral acceleration
#thoracic spine
AcclML_brms = brm(Accl_ML_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                  chains = 4, iter = 4000,
                  control = list(max_treedepth = 15),
                  data = dplyr::filter(DataGPSStep, PlayerID %in% Players_for_Val),
                  prior = AcclML_priors)
```

```
summary(AcclML_brms)

#thoracic spine
AcclML_brms_lum = brm(Accl_ML_av ~ Group * Activity * Days_Since_Conc + (1|PlayerID),
                  chains = 4, iter = 4000,
                  control = list(max_treedepth = 15),
                  data = dplyr::filter(Data_LumbarStep, PlayerID %in% Players_for_Val),
                  prior = AcclML_priors)

summary(AcclML_brms_lum)

#output for the models
AcclML_brms_output = bind_rows(summary(AcclML_brms)$fixed, summary(AcclML_brms_lum)$fixed)
%>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ),
         Term = gsub("[...]", "", Term),
         Term = gsub("[0-9]", "", Term),
         Location = if_else(row_number()<=12, "Thoracic", "Lumbar") )%>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Location)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Mediolateral acceleration (g)', .before = 1)


#hypothesis test for MLical acceleraiton
AcclML_hypothesis = bind_rows(hypothesis(AcclML_brms, c("GroupConcussed > 0"))$hypothesis,
                                hypothesis(AcclML_brms, c("GroupConcussed:ActivityJog > 0")
)$hypothesis,
                                hypothesis(AcclML_brms, c("GroupConcussed:ActivityRun > 0")
)$hypothesis,
                                hypothesis(AcclML_brms, c("GroupConcussed:Days_Since_Conc <
0"))$hypothesis,
                                hypothesis(AcclML_brms, c("GroupConcussed:ActivityJog:Days_
Since_Conc + GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
                                hypothesis(AcclML_brms, c("GroupConcussed:ActivityRun:Days_
Since_Conc + GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed > 0"))$hypoth
esis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed:ActivityJog >
0"))$hypothesis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed:ActivityRun >
0"))$hypothesis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed:Days_Since_Co
nc < 0"))$hypothesis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed:ActivityJog:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc < 0"))$hypothesis,
                                hypothesis(AcclML_brms_lum, c("GroupConcussed:ActivityRun:D
ays_Since_Conc + GroupConcussed:Days_Since_Conc < 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2),
         Location = if_else(row_number()<=6, "Thoracic", "Lumbar"))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))



####contact time
#these are weak informative priors with a flat distribution
#set priors
ConTime_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0.1, 0.1)", class = "b", coef = "GroupConcussed"), # Prior for GroupCon
```

```
cussed effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(0.05, 0.05)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prio
r for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#run model
ConTime_brms = brm(CT_apex_all_av ~ Group * Activity * Days_Since_Conc + SpeedBF_av +(1|Pla
yerID),
                   chains = 4, iter = 4000,
                   control = list(max_treedepth = 15),
                 data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val & Activity !=
"Walk"),
                 prior = ConTime_priors)

summary(ConTime_brms)

#output for contact time
ConTime_brms_output = summary(ConTime_brms)$fixed %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ))%
>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Tail_ESS)%>%
  rename(`Bulk ESS` = Bulk_ESS,
        `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Contact time (s)', .before = 1)

#hypothesis test for ML acceleraiton
ConTime_hypothesis = bind_rows(hypothesis(ConTime_brms, c("GroupConcussed > 0"))$hypothesis
,
                              hypothesis(ConTime_brms, c("GroupConcussed:ActivityRun > 0"))
$hypothesis,
                              hypothesis(ConTime_brms, c("GroupConcussed:Days_Since_Conc  +
GroupConcussed:Days_Since_Conc< 0"))$hypothesis,
                              hypothesis(ConTime_brms, c("GroupConcussed:ActivityRun:Days_S
ince_Conc  + GroupConcussed:Days_Since_Conc< 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))



####VRGF time
#set priors
vGRF_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0, 0.3)", class = "b", coef = "GroupConcussed"), # Prior for GroupConcu
ssed effect
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(0, 0.3)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior fo
r GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
```

```
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#vGRF model
vGRF_brms = brm(vGRF_apex_all_av ~ Group * Activity * Days_Since_Conc + SpeedBF_av +(1|Play
erID),
                chains = 4, iter = 4000,
                control = list(max_treedepth = 15),
                   data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val & Activity !
= "Walk"),
                prior = vGRF_priors)

summary(vGRF_brms)

vGRF_brms_output = summary(vGRF_brms)$fixed %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ))%
>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Tail_ESS)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Vertical ground reaction force (kN)', .before = 1)

#hypothesis test for vRGF
vGRF_hypothesis = bind_rows(hypothesis(vGRF_brms, c("GroupConcussed < 0"))$hypothesis,
                               hypothesis(vGRF_brms, c("GroupConcussed:ActivityRun < 0"))$h
ypothesis,
                               hypothesis(vGRF_brms, c("GroupConcussed:Days_Since_Conc  + G
roupConcussed:Days_Since_Conc> 0"))$hypothesis,
                               hypothesis(vGRF_brms, c("GroupConcussed:ActivityRun:Days_Sin
ce_Conc  + GroupConcussed:Days_Since_Conc> 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

####contact time
DataGPSStep %>%
  group_by(Group, Activity)%>%
  dplyr::filter(Assessment == 1  & PlayerID %in% Players_for_Val)%>%
  summarise(mean = mean(COMdis_apex_all_mean_av, na.rm = T),
            sd = sd(COMdis_apex_all_mean_av, na.rm = T))%>%
  pivot_wider(values_from = c(mean:sd), names_from = Group)%>%
  mutate(diff = mean_Concussed-mean_Control,
         ES = (mean_Concussed-mean_Control)/(sd_Control+sd_Concussed/2))

#set priors
COM_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(-0.5, 0.05)", class = "b", coef = "GroupConcussed"), # Prior for GroupC
oncussed effect
  set_prior("normal(0.2, 0.05)", class = "b", coef = "ActivityRun"), # Prior for ActivityRu
n effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(-0.5, 0.05)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prio
r for GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
```

```
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#run the model
COM_brms = brm(COMdis_apex_all_mean_av ~ Group * Activity * Days_Since_Conc + SpeedBF_av +
(1|PlayerID),
               chains = 4, iter = 4000,
               control = list(max_treedepth = 15),
                data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val & Activity != "
Walk"),
               prior = COM_priors)
summary(COM_brms)

COM_brms_2 = brm(COMdis_apex_all_mean_av ~ Group * Activity * Days_Since_Conc + SpeedBF_av
+ (1|PlayerID),
                  chains = 4, iter = 4000,
                  control = list(max_treedepth = 15),
                  data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val & Activity !=
"Walk"),
                 prior = COM_priors)

COM_brms_output = summary(COM_brms_2)$fixed %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ))%
>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Tail_ESS)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Centre of mass displacement (cm)', .before = 1)

#hypothesis test for vRGF
COM_hypothesis = bind_rows(hypothesis(COM_brms_2, c("GroupConcussed < 0"))$hypothesis,
                           hypothesis(COM_brms_2, c("GroupConcussed:ActivityRun < 0"))$h
ypothesis,
                           hypothesis(COM_brms_2, c("GroupConcussed:Days_Since_Conc  + G
roupConcussed:Days_Since_Conc> 0"))$hypothesis,
                           hypothesis(COM_brms_2, c("GroupConcussed:ActivityRun:Days_Sin
ce_Conc  + GroupConcussed:Days_Since_Conc> 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))

###flight time model
DataGPSStep %>%
  group_by(Group, Activity)%>%
  dplyr::filter(Assessment == 1)%>%
  summarise(mean = mean(FT_apex_all_av, na.rm = T),
            sd = sd(FT_apex_all_av, na.rm = T))%>%
  pivot_wider(values_from = c(mean:sd), names_from = Group)%>%
  mutate(diff = mean_Concussed-mean_Control,
         ES = (mean_Concussed-mean_Control)/(sd_Control+sd_Concussed/2))

#set priors
FT_priors = c(
  set_prior("", class = "Intercept"), # Prior for the intercept
  set_prior("normal(0, 0.3)", class = "b", coef = "GroupConcussed"), # Prior for GroupConcu
ssed effect
```

```
  set_prior("", class = "b", coef = "ActivityRun"), # Prior for ActivityRun effect
  set_prior("", class = "b", coef = "Days_Since_Conc"), # Flat prior for Days_Since_Conc ef
fect
  set_prior("normal(0, 0.3)", class = "b", coef = "GroupConcussed:ActivityRun"), # Prior fo
r GroupConcussed:ActivityRun interaction
  set_prior("", class = "b", coef = "GroupConcussed:Days_Since_Conc"), # Flat prior for Gro
upConcussed:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "ActivityRun:Days_Since_Conc"), # Flat prior for Activi
tyRun:Days_Since_Conc interaction
  set_prior("", class = "b", coef = "GroupConcussed:ActivityRun:Days_Since_Conc") # Flat pr
ior for GroupConcussed:ActivityRun:Days_Since_Conc interaction
)

#run the model
FT_brms = brm(FT_apex_all_av ~ Group * Activity * Days_Since_Conc + SpeedBF_av + (1|PlayerI
D),
              chains = 4, iter = 4000,
              control = list(max_treedepth = 15),
              data = dplyr::filter(DataGPS, PlayerID %in% Players_for_Val & Activity != "W
alk"),
              prior = FT_priors)

FT_brms_output = summary(FT_brms)$fixed %>%
  rownames_to_column('Term')%>%
  mutate(across(c(Estimate:`u-95% CI`), ~ round(., 3)))%>%
  mutate(across(c(Bulk_ESS:Tail_ESS), ~ round(., 0)))%>%
  mutate(`Estimate [95% CI]` = paste0(Estimate, " [",`l-95% CI`, " to ", `u-95% CI`,"]" ))%
>%
  dplyr::select(Term, `Estimate [95% CI]`, Bulk_ESS:Tail_ESS)%>%
  rename(`Bulk ESS` = Bulk_ESS,
         `Tail ESS` = Tail_ESS)%>%
  add_row(Term = 'Centre of mass displacement (cm)', .before = 1)

#hypothesis test for vRGF
FT_hypothesis = bind_rows(hypothesis(FT_brms, c("GroupConcussed < 0"))$hypothesis,
                          hypothesis(FT_brms, c("GroupConcussed:ActivityRun < 0"))$hypothe
sis,
                          hypothesis(FT_brms, c("GroupConcussed:Days_Since_Conc + GroupCon
cussed:Days_Since_Conc > 0"))$hypothesis,
                          hypothesis(FT_brms, c("GroupConcussed:ActivityRun:Days_Since_Con
c + GroupConcussed:Days_Since_Conc > 0"))$hypothesis)%>%
  dplyr::select(1,6:7)%>%
  mutate(`Bayes factor` = round(Evid.Ratio, 2),
         `Posterior probability` = round(Post.Prob, 2))%>%
  dplyr::select(-c(Evid.Ratio, Post.Prob))
```

An example of the calculation of test re-test reliabilty for one of the outcome variables, this was repeated for each variable measured

```
icc_result = data.frame(ICC(Rel_Data_Steps %>%
                                dplyr::filter(Activity == paste(Current_Activity) & Pace ==
paste(Current_Pace))%>%
                                dplyr::select(Cadence_1:Cadence_5))$results[3,],
                          row.names = F)


  #calculate SD from first session
  sd = Rel_Data_Steps %>% ungroup() %>%
    dplyr::filter(Activity == paste(Current_Activity) & Pace == paste(Current_Pace)) %>%
    dplyr::select(Activity, Pace, Cadence_1)%>%
    pivot_longer(Cadence_1, values_to = "Value", names_to = "Variable")%>%
    summarise(sd = sd(Value, na.rm = T))

  #calculate the mean for each activity
```

```r
mean = Rel_Data_Steps %>% ungroup() %>%
  dplyr::filter(Activity == paste(Current_Activity) & Pace == paste(Current_Pace)) %>%
  dplyr::select(Activity, Pace, Cadence_1:Cadence_5)%>%
  pivot_longer(Cadence_1:Cadence_5, values_to = "Value", names_to = "Variable")%>%
summarise(mean = mean(Value, na.rm = T))

#calculate the max for each activity
max = Rel_Data_Steps %>% ungroup() %>%
  dplyr::filter(Activity == paste(Current_Activity) & Pace == paste(Current_Pace)) %>%
  group_by(PlayerID)%>%
  dplyr::select(Activity, Pace, Cadence_1:Cadence_5)%>%
  pivot_longer(Cadence_1:Cadence_5, values_to = "Value", names_to = "Variable")%>%
  na.omit()%>%
  summarise(max = max(Value, na.rm = T))%>%
  ungroup()%>%
  summarise(max = mean(max, na.rm = T))

icc_df = cbind(icc_result, max, mean, sd)%>%
  mutate(SEM = round(sd*sqrt(1-ICC),8),
         MDC = round(1.96 * SEM * sqrt(2),8),
         Rel_MDC = MDC/mean*100,
         Current_Activity,
         Current_Pace,
         Variable = "Cadence",
         Location = Current_Location)
```