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An Exploratory Study of the Acute Effects of Football Heading on Postural Control and Corticospinal Inhibition

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Abstract

Repetitive non-concussive head impacts (NCHIs) may contribute to long-term neurodegenerative conditions. However, objective, multimodal methods for monitoring acute changes in brain health biomarkers following NCHIs remain underdeveloped. In this exploratory study, we examined the effects of ten kicking and ten heading trials related to association football on linear and nonlinear measures of postural control and corticospinal inhibition. Postural control was assessed via force platform analysis in dual-stance and single-leg protocols, and corticospinal inhibition was measured using transcranial magnetic stimulation with electromyography. Large effects of condition were found for anteroposterior postural complexity (CI-AP), anteroposterior sway amplitude, mediolateral centre of pressure shift and cortical silent period ($\eta^2 > 0.14$). Pairwise comparisons revealed large post-heading effects, particularly in CI-AP, which decreased significantly relative to baseline ($d_z = 0.71$, $p = 0.018$) and showed a moderate negative effect relative to post-kicking testing ($d_z = 0.53$, $p = 0.069$). These findings suggest a possible reduction in postural control adaptability following exposure to ten NCHIs, consistent with patterns observed in mild traumatic brain injury. Whilst confirmatory research with larger samples is warranted, nonlinear measures of postural control complexity demonstrate promise as a sensitive biomarker for detecting acute NCHI-related changes.

Keywords: non-concussive head impacts; postural control; complexity; transcranial magnetic stimulation; football heading

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1. Introduction

Growing evidence suggests that repetitive, non-concussive head impacts (NCHIs) in contact and collision sports may negatively affect long-term brain health [1–3]. Though not associated with concussion-like symptoms [4], laboratory research has shown that an accumulation of NCHIs can acutely impair brain neurophysiology [5,6], cognition [7], and postural control [8–10]. Moreover, retrospective analyses have suggested a possible association between chronic exposure to NCHIs and an increased risk of developing neurodegenerative conditions such as Alzheimer's disease and chronic traumatic encephalopathy [1,3,11]. As such, the medical community has been prompted to assume a causal relationship between repetitive NCHI exposure and risk of neurological disease development, while researchers have been called to develop monitoring strategies to provide a more evidence-based understanding of how NCHIs may impair brain health [3]. Currently, biochemical sampling and neuroimaging are among the most robust methods for determining subtle brain injury [10], but high cost, time requirements, and specialist expertise limit their use in the field [12]. Therefore, there is a need to develop accessible, quantitative assessment strategies that can reliably identify markers of NCHI-related brain injury within athletic environments.

Postural control is a fundamental aspect of human movement, reflecting the ability to maintain equilibrium and stability during both standing and dynamic tasks [13]. Healthy postural control relies on inputs from the vestibular, visual and somatosensory systems [14], which are often disrupted following mild traumatic brain injury [15]. As a result, postural control deficits are a cardinal symptom of sports-related concussion (SRC) [16], evident in approximately 60% of SRC cases [17]. Importantly, athletes may still exhibit underlying postural control deficits even after being cleared for return-to-sport, which may contribute to the twofold increase in musculoskeletal injury risk among post-SRC athletes [18]. Given that the postural control system is highly vulnerable to neural disruption [14], the increased musculoskeletal injury risk observed post-SRC may also extend to athletes frequently exposed to repetitive NCHIs, underscoring the value of monitoring postural control in this population. Notably, postural control assessments are generally non-invasive, portable, and adaptable to field environments.

Current approaches for postural control assessment in SRC-related cases include both observational and instrumented assessment protocols [19]. In field settings, the most widely adopted tool is the Modified Balance Error Scoring System (mBESS)—a component of the Sport Concussion Assessment Tool [20]. Paper-based, the mBESS scores performance by a count of losses in balance across double-, single-, and tandem-stance trials [21]. Though easy to administer, limitations of the mBESS include its reliance on subjective error scoring, poor test–retest reliability, and limited construct validity [16,22]. In contrast, instrumented protocols involving force platforms can offer more objective and reliable measurements, capturing temporal changes in centre of pressure (COP) position [23]. In clinical practice, the Sensory Organisation Test (SOT) represents the gold standard in objective postural control assessment [24,25], utilising a sway-referenced force platform and visual surroundings to evaluate sensory contributions to postural control [26]. While effective at identifying acute deficits post-SRC [24], the high cost and immobility of the SOT restrict its use to advanced clinical environments [25]. Furthermore, its utility as a recovery-monitoring tool has been questioned, as SOT metrics often normalise within three days post-SRC [27], whereas nonlinear COP measures can identify postural control deficits at up to 21 days post-SRC incidence [19]. Together, these findings highlight the limitations of current approaches and the need for objective, field-transferable methods, therefore motivating interest in nonlinear approaches to COP analysis.

While research has traditionally emphasised linear COP measures such as sway velocity, path length, and amplitude [28], these metrics can lack sensitivity to subtle neural disruptions [19,29]. Nonlinear approaches, particularly measures of postural control complexity, provide a richer characterisation of sway by quantifying the structure and variability of the time series data [30]. Complexity reflects the integration of sensory, vestibular, cognitive, and motor processes, as well as the system's adaptability to environmental demands [19]. According to the loss of complexity hypothesis, physiological and motor systems naturally operate across multiple time scales, and reductions in complexity signify a diminished adaptability to stressors and changing environments [31]. Consistent with this hypothesis, reductions in postural complexity—reflected by more rigid and predictable sway patterns—are commonly observed in pathological populations [32] and in athletes recovering from SRC [33,34]. Notably, such changes can be evident despite linear measures remaining unchanged [33] or having already resolved [19]. Postural complexity may therefore offer superior sensitivity as a biomarker of subtle postural control deficits, particularly in the context of NCHI exposure [35].

Association Football (AF) is the most popular team sport worldwide [36], uniquely exposing players to deliberate, routine NCHIs via "heading" [37]. While the social, physical, and mental benefits of AF participation are well established [38,39], growing evidence has associated chronic heading performance with the onset of neurodegenerative conditions later in life [1,11,40]. Notably, a 3.5-fold greater risk of neurodegenerative disease mortality was reported among retired AF players in Scotland, relative to matched controls [40]. This finding led the Scottish Football Association to ban heading practice on the days immediately before and after matches [41,42]. Similarly, the English Football Association has acted to prohibit heading among players under 12 years of age [43,44], and published training guidance for adult players, recommending a maximum of one session per week of no more than 10 headers [45]. These policies reflect a growing awareness of NCHI exposure and the urgent need for evidence-based monitoring strategies to mitigate potential long-term brain injury risk.

Previous investigations into the possible relationship between AF heading and brain injury development have employed a range of vestibular, neurocognitive and neurophysiological assessments adapted from concussion research, most often focusing on linear measures of postural control [5,8,9,46–52]. In some studies, exposures of 10–20 headers have been linked to acute changes in brain injury markers, though findings remain mixed. Far less is known about the effects of NCHIs on nonlinear postural control measures, particularly complexity. For example, a recent study examining the acute effects of 12 headers reported increased sway velocity post-heading, but no change in approximate entropy—a measure of sway regularity [35]. Importantly, emerging evidence suggests that multiscale entropy may provide a richer characterisation than single-scale entropies, potentially offering greater sensitivity for analysing postural control regulation [53]. Additionally, while cortical silent period (cSP)—a neurophysiological measure of corticospinal inhibition—has shown promise as a biomarker of concussion [54,55], its sensitivity to non-concussive injury remains unclear. Although Di Virgilio et al. [5] reported measurable increases in cSP following heading, more recent evidence suggests that comparable changes may also occur in response to tasks without head impacts [49]. Therefore, this exploratory study investigated the acute effects of ten kicking and ten heading trials on linear and nonlinear postural control measures and corticospinal inhibition, aiming to identify potential markers of NCHI for future confirmatory research.

2. Materials and Methods

2.1. Participants

Nineteen males (mean $\pm$ SD; age 26.9 $\pm$ 8.4 years, mass 76.8 $\pm$ 14.7 kg, height 1.79 $\pm$ 0.08 m) with at least two years of previous AF heading training volunteered to take part in this study. Prior to experimental participation, participants completed a specifically designed pre-exercise health/medical questionnaire to screen for musculoskeletal/facial injuries, concussion incidence in the previous six months, and neurological conditions that may have contraindicated participation or influenced study outcomes. Participants also completed a specifically designed TMS-focused screening questionnaire, to determine their suitability for TMS exposure. No contraindications for participation or experimental findings were reported. Participants were asked to avoid strenuous exercise within two hours of their session start time.

All experimental procedures were conducted in accordance with the principles outlined in the Declaration of Helsinki (1975, revised in 2013) and were approved by the Sheffield Hallam University Human Research Ethics Committee (ER44471254) on 3 November 2022. Reciprocal ethics approval was also provided by the La Trobe University Human Research Ethics Committee. Written informed consent was obtained from all participants prior to their involvement in the study.

2.2. Sample Size Estimation and Justification

Resource and study timescale constraints restricted the achievable sample size to 19 participants, and the study accordingly used an exploratory design. A sensitivity power analysis for a repeated-measures ANOVA (within factors) was conducted using G*Power (V 3.1.9.6, Keil, Germany). Data loss of the key outcome variable (anteroposterior complexity index, CI-AP) due to signal artefact reduced the sample of this measure to 13. Therefore, with 13 samples, an α error probability of 0.05, and a correlation of 0.72 between measures of CI-AP, 80% power would be achieved for $f > 0.280$ ($\eta^2 = 0.07$). A sensitivity power analysis was also performed for a t test (matched pairs). With 13 samples, an α error probability of 0.05, 80% power would be achieved for $d_z \geq 0.85$.

2.3. Study Design

This study incorporated a within-participant design to investigate the potential acute changes in brain health measures following a kicking (control) and heading (exposure) activity. Thus, participants were examined during three testing conditions: baseline, post-kicking, and post-heading. Chronologically, the testing battery included assessment of corticospinal inhibition using transcranial magnetic stimulation (TMS), dual-stance postural control with a concurrent Stroop test dual-task, and single- and double-legged balance performance using the commercially available Sparta Science™ Balance Scan (hereafter referred to as the "[Sparta] Balance Scan").

Prior to baseline testing, participants completed one familiarity trial of the dual-stance postural control and Balance Scan assessments, intended to minimise the potential learning effects of re-testing. Familiarity trials of the TMS protocol were not prescribed, as neurophysiological responses to electromagnetic stimuli are involuntary. To mitigate the risk of musculoskeletal injury in the kicking and heading interventions, participants were then prescribed a standardised warm-up consisting of mobility and coordination exercises. The lower-body component included ten lateral lunges, ten reverse lunges, dynamic leg swings, and hip internal/external rotations ("open and close the gates"). Neck mobility exercises included three lateral flexions, three oblique neck extensions, and two slow half-neck rolls. Finally, a short coordination task involved five overhead throws and catches.

A flowchart demonstrating the experimental order of events is presented in Figure 1. A size 5 Nike Flight soccer ball was inflated to 11 psi (the minimum recommended inflation by the manufacturer) at the beginning of each experimental session and utilised for all kicking and heading trials. Each participant completed the entire experimental session (including rest periods) within two hours.

2.4. Kicking Task

A 10-trial kicking task was prescribed as a control intervention. For each trial, a researcher rolled the ball to the participant, instructing them to take a preparatory touch before shooting toward a 1 × 1.2 m target net positioned ten metres away (Figure 2). To ensure consistency with the subsequent heading activity, a 60 s break period was maintained between trials.

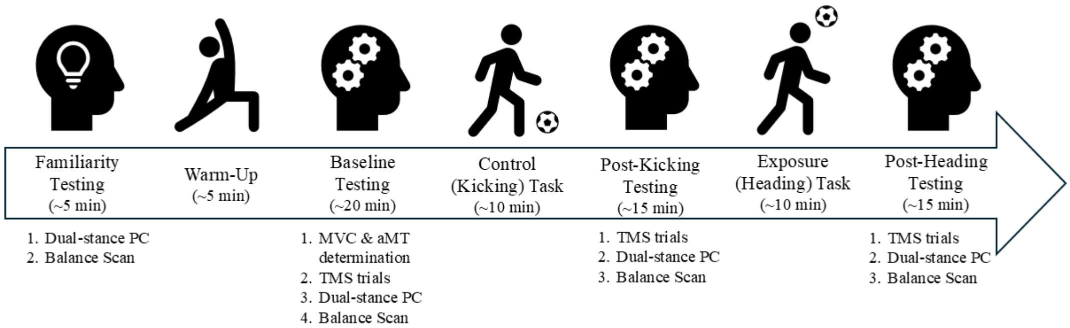

Figure 1. Experimental session order and timings of events. Numbered lists demonstrate the assessment order at each stage. Assessments occurred moments after one another. MVC and aMT determination relates to preparatory TMS protocols outlined in Section 2.6 Corticospinal Inhibition. Abbreviations: aMT, active motor threshold; MVC, maximal voluntary contraction; PC, postural control; TMS, transcranial magnetic stimulation.

Figure 2. The kicking task involved ten "take a touch and shoot" trials toward a 1 × 1.2 m target net positioned ten metres away.

2.5. Heading Task

In line with the English Football Association Adult Amateur Heading Guidance (England Football Association, 2021a [45]), participants were prescribed ten headers, and the ball was delivered from a 10 m distance. Initially, a ball launcher (Eurogoal 1500, GLOBUS, Codognè, Italy) was utilised to standardise delivery velocity. However, following mechanical failure after the first participant, manual overhead throws were adopted (Figure 3), simulating throw-ins and examples from previous research [48,56].

As with the control intervention, a minimum 60 s rest interval was provided following each heading trial. To promote technique consistency, participants were instructed to step toward the incoming ball and head it back towards the researcher without jumping [8,50]. The participants completed several practice trials (catching the ball overhead) to familiarise themselves with the ball delivery before heading.

Peak resultant head accelerations during the heading trials were measured using a pre-fitted instrumented mouthpiece (v1.4, Prevent Biometrics, Edina, MN, USA) equipped with a triaxial accelerometer. Heading trials were also recorded using 2D video in the sagittal plane, allowing for *post hoc* measurement of ball delivery velocity (11.43 ± 0.77 m/s; mean ± SD) using video annotation software (Kinovea, v.9.5). Average ball delivery velocity in the current study remained comparable to velocities of 10.75–15.56 m/s reported in prior research [5,9,47,49,50].

Figure 3. The heading task involved ten frontal header trials of a size 5 ball, delivered by overhead throw from 10 m range.

2.6. Corticospinal Inhibition

Following previously described methods [49,54], TMS was applied over the optimal site of the contralateral motor cortex targeting the first dorsal interosseous (FDI) muscle of the participants' dominant hand. The skin surface at the FDI was prepared with a gauze swab [57] before a Delsys Trigno wireless electromyography (EMG, 2148 Hz) sensor was attached over the FDI muscle belly.

A Magstim 200^2 stimulator with a 90 mm circular coil (Magstim Co., Whitland, UK) was used to generate single-pulse stimuli. The optimal stimulation site was determined by observing the largest discernible active motor evoked potential (MEP) through exploration of the estimated area of hand muscles [58]. For reliable coil placement, participants wore a tightly fitted cap (EasyCap, Herrsching, Germany), positioned with reference to the nasion-inion and inter-aural lines to reliably locate the optimal site over the motor cortex for repeated testing. A customised wooden board was designed to assist with stabilising the wrist and finger digits 3–5 for controlled assessment of the FDI (Figure 4). The board included a socket for an electronic hanging scale (with continuous digital readout), which was used as a force transducer. Maximal voluntary contraction (MVC) of the FDI was determined from three finger abduction trials against the scale. Active motor threshold (aMT) was then quantified during abduction of the first finger at 10% MVC. Determination of aMT was achieved through observation of an MEP at the lowest TMS intensity in three of five stimuli at the optimal site [59]. After determination of the aMT, 20 stimuli were conducted (in four sets of five trials) at 130% aMT. Each stimulus was triggered at random 4–6 s intervals to minimise anticipatory effects, while the participant maintained FDI abduction at 10% MVC.

After the session, EMG data were plotted using MATLAB (version R2023a; MathWorks, Natick, MA, USA) and the cortical silent period (cSP) of each trial was manually recorded from the MEP onset to the resumption of uninterrupted EMG activity (milliseconds, ms), representing corticospinal inhibition [5,54,60]. MEP amplitudes were also recorded from maximum to minimum peaks (millivolts, mV) for each trial.

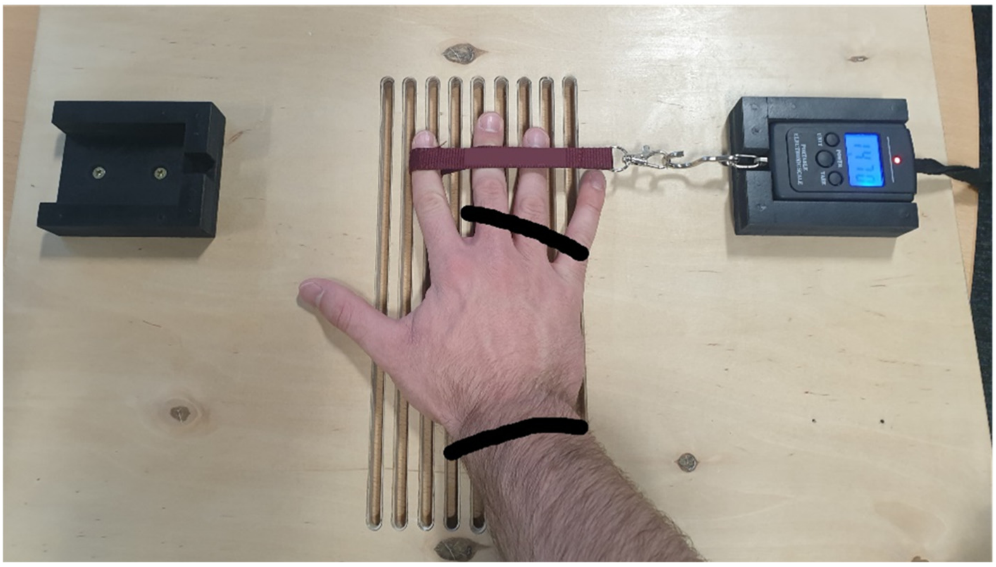

Figure 4. A customised board with fixed sockets for the digital hanging scale (proxy force transducer). The first digit is seen abducting, creating tension on the lanyard and producing a force reading. Hook-and-loop straps tied to the air vent enabled securement of the wrist and digits 3–5.

2.7. Dual-Stance Postural Control Assessment

Postural control assessments with accompanying dual tasks have been previously utilised to incorporate an external focus of attention and elicit more natural postural responses [61–64]. In the current study, participants stood with feet shoulder-width apart [64] for 60 s atop a force platform (BMS600900, AMTI, Watertown, MA, USA), which was embedded beneath 3 mm thick vinyl flooring and sampled at 1000 Hz. Participants were instructed to maintain a still, upright dual stance throughout the trial while completing a Stroop test. The Stroop test involved the presentation of 60 incongruent colour-word stimuli presented at one-second intervals. Participants were instructed to name the font colour aloud as accurately as possible. The Stroop test was displayed on a 46″ screen positioned at eye-level from a 2 m distance.

The force platform captured force and moment data in the mediolateral (ML, x), anteroposterior (AP, y) and vertical (z) axes. Consistent with previous research [19], force and moment data were downsampled to 100 Hz (6000 samples per trial) to reduce computational load and minimise data redundancy. This sampling rate is sufficient to preserve signal integrity, as most postural sway during dual stance occurs at frequencies well below 10 Hz [65]. A fast Fourier transform was conducted, informing the decision to filter the signal using a zero-lag 2nd-order Butterworth filter with a 10 Hz cut-off.

Centre of pressure (COP) position in the ML and AP axes was calculated from force and moment data [66] in MATLAB and subsequently used for linear and nonlinear postural control measurement. Linear measures included the root mean square of the instantaneous COP displacements from the mean position in the ML and AP axes (RMS-ML, RMS-AP)—representing sway amplitudes in those directions, COP path length, mean COP velocity and COP sway area (95% confidence ellipse). For nonlinear assessment, multiscale entropy was calculated using the following parameters: $m = 2$, $r = 0.15 \times SD$, and timescales = 30. Complexity index (CI) was then calculated as the sum of sample entropies across each timescale, determining the area under the multiscale entropy curve. Complexity indices were calculated in both the ML (CI-ML) and AP (CI-AP) directions.

2.8. Sparta Balance Scan

A commercially available balance assessment tool, the Sparta Science™ (since acquired by Oura™) Balance Scan, was incorporated as a portable, field-accessible alternative in postural control assessment. The Balance Scan utilises a proprietary force platform (SSFP01, Menlo Park, CA, USA) and associated Sparta Scan software (version 1.2.6), to examine single-leg and dual-stance balance performance. The Balance Scan protocol guided participants through an initial weigh-in (2 s), before alternating intervals of left and right single-leg (8 s) and dual-stance (6 s) for two cycles. In line with standard Balance Scan protocols, participants were presented with live COP trace feedback to self-regulate balance. The trace feedback was displayed on a 46″ screen positioned at eye level from a 2 m distance. The Balance Scan protocol spanned approximately 56 s, with the Scan capturing two seconds of static dual-stance during the weigh-in, and five seconds of static single-leg stance during each single-leg interval.

After the session, the raw Balance Scan dataset of 68 metrics was exported to Excel. Left and right single-leg data were manually reorganised into 'kicking' and 'non-kicking' leg variables, according to each participant's preferred kicking leg. To reduce the dataset to non-correlated variables, a principal component analysis (PCA; varimax rotation) was conducted after standardising data to z-scores. The initial 68-metric dataset failed Barlett's Test of Sphericity, though sphericity was achieved (χ^2 = 3108.96, df = 378, $p < 0.001$) with a Kaiser-Meyer-Olkin measure of 0.50 following the removal of body weight, body mass, t-score and symmetry-based metrics. This resulted in 31 Balance Scan metrics being entered into the final PCA. Subsequently, components 1–7 were retained based on an eigenvalue threshold of 1 [67], 90% cumulative variance [68], and inspection of the scree plot [69]. Thereafter, only variables with a factor loading >0.7 were considered for inclusion [68,70,71], and a single variable was retained from each component based on maximum factor loading or relevance to the study's aims [70]. Subsequently, seven Balance Scan metrics were selected for further analysis. In order of components 1–7, these variables were kicking leg balance, multiscale vertical entropy (MVE), non-kicking leg balance, COPx shift (m), multivariate multiscale entropy in the XY Plane (mMSE-XY), multivariate multiscale entropy during dual stance (mMSE-2), and multivariate multiscale entropy in the XY plane during dual stance (mMSE-XY-2). Definitions of each selected Sparta Balance variable are presented in Table 1. The Sparta system utilised mMSE as its default entropy measure, differing from the more common univariate MSE by analysing COP dynamics in the mediolateral and anteroposterior axes simultaneously [72].

Table 1. Sparta Balance Scan variable definitions.

Variable Name	Definition
Kicking leg/non-kicking leg balance	A proprietary Sparta metric (0–100) representing single-leg postural stability, derived from sway velocity and related measures.
COPx shift (m)	The shift in x-axis (mediolateral) centre of pressure when moving from two legs to one. Measured in m.
Multiscale vertical entropy [MVE]	A multiscale sample entropy measurement for the vertical force signal over the target window * (5 s per single-leg stance). Measured in arbitrary units.
Multivariate multiscale entropy in XY plane [mMSE-XY]	A multivariate, multiscale sample entropy measurement for centre of pressure (x-y axes) and vertical force over the single-leg target windows. Measured in arbitrary units.

Table 1. *Cont.*

Variable Name	Definition
Multivariate multiscale entropy—2 leg [mMSE-2]	A multivariate, multiscale sample entropy measurement for centre of pressure (x-y axes) and vertical force over the weigh-in target window (2 s). Measured in arbitrary units.
Multivariate multiscale entropy in XY plane—2 leg [mMSE-XY-2]	A multivariate, multiscale sample entropy measurement for centre of pressure (x-y axes) over the weigh-in target window. Measured in arbitrary units.

* The target window refers to the portion of each stance interval where recorded data is utilised for computation of that metric. For single-leg metrics, the target window is 5 S per single leg interval, and for 2—leg metrics, the target window is 2 S during the initial weigh-in.

2.9. Statistical Analysis

Acute, within-participant responses to the kicking and heading interventions were analysed using linear mixed-effects models via the *gamlj* module in jamovi (version 2.3.17.0). All outcome variables were modelled as a function of Condition, with data clustered within Participant. For most models, a random intercept for Participant and a random slope for Condition were included to account for individual variability in condition-related effects (i.e., 'Variable_Name' ~ 1 + Condition + (1 + Condition | Participant)). The exception was cSP, which included MEP amplitude as both a fixed effect and a random slope: cSP ~ 1 + Condition + MEP amplitude + (1 + Condition + MEP amplitude | Participant).

Eta-squared was calculated to quantify the effect of condition on the assessed variables, interpreted according to thresholds of η^2 = 0.01 (small), 0.06 (medium) and 0.14 (large) [73]. Between-condition *post hoc* pairwise comparisons were quantified using Cohen's d_z effect sizes, calculated using the formula provided by Rosenthal (1991) [74]: $Cohen's\ d_z = \frac{t}{\sqrt{n}}$. Cohen's d_z values were interpreted as negligible ($\leq$0.19), small (0.20–0.49), moderate (0.50–0.79), and large ($\geq$0.80) effects, adapting conventional thresholds by Cohen (1988) [75] to provide additional granularity for effects below 0.20. Additionally, post hoc p values are reported uncorrected, reflecting the exploratory aim of the study [76]. Readers should interpret significance cautiously, with emphasis on effect sizes and confidence intervals rather than p values alone.

3. Results

Data from the instrumented mouthpiece showed that participants headed the ball at an average peak linear acceleration of 16.33 ± 2.49 g (mean ± SD) and an average peak angular acceleration of 1016 ± 296 rad/s2. Of the 19 participants examined, one declined TMS, and EMG data from two participants were excluded due to poor signal quality. Post-kicking and post-heading EMG data from one of these participants were retained. Additional data loss occurred due to signal artefact in dual-stance postural control data in seven participants, an iMG sync error for head acceleration data in one participant, and non-capture of sagittal plane recordings of ball delivery in four participants. Estimated marginal means and repeated-measures omnibus test results are summarised in Table 2, with pairwise differences presented in Table 3. In the text, all differences are described in terms of their direction (increase/decrease), while t-value signs reflect the subtraction order used by jamovi. All standardised within-subject effect sizes (d_z) were calculated at the participant level.

Table 2. Linear Mixed Model results of linear and nonlinear measures of dual-stance postural control and Balance Scan performance, and corticospinal inhibition across experimental conditions. Data are estimated marginal mean (EMM) ± standard error (SE) and 95% confidence intervals [95% CI].

	Condition			ANOVA			
Variable	**Baseline**	**Post-Kicking**	**Post-Heading**	F	**df**	p	η^2
Dual-stance postural control nonlinear measures							
CI-ML							
EMM + SE	31.49 ± 3.80	29.29 ± 3.80	30.84 ± 3.80	0.15	2, 24	0.863	0.01
[95% CI]	[23.69–39.28]	[21.49–37.08]	[23.04–38.63]				
n	13	13	13				
CI-AP							
EMM + SE	25.54 ± 1.88	24.33 ± 1.88	20.79 ± 1.88	3.52	2, 24	0.046 *	0.23
[95% CI]	[21.65–29.42]	[20.45–28.22]	[16.90–24.67]				
n	13	13	13				
Dual-stance postural control linear measures							
RMS-ML (mm)							
EMM + SE	3.42 ± 0.89	4.17 ± 0.89	3.28 ± 0.89	0.54	2, 24	0.591	0.04
[95% CI]	[1.60–5.25]	[2.35–6.00]	[1.45–5.11]				
n	13	13	13				
RMS-AP (mm)							
EMM + SE	5.58 ± 0.87	5.64 ± 0.87	7.77 ± 0.87	3.00	2, 24	0.069	0.20
[95% CI]	[3.80–7.37]	[3.86–7.43]	[5.98–9.55]				
n	13	13	13				
COP Path Length (mm)							
EMM + SE	876.69 ± 75.07	886.51 ± 75.07	845.55 ± 75.07	0.19	2, 24	0.828	0.02
[95% CI]	[720.89–1032.49]	[730.71–1042.31]	[689.75–1001.35]				
n	13	13	13				
Mean COP Velocity (mm/s)							
EMM + SE	14.61 ± 1.25	14.77 ± 1.25	14.12 ± 1.25	0.17	2, 24	0.846	0.01
[95% CI]	[12.01–17.21]	[12.17–17.37]	[11.53–16.72]				
n	13	13	13				
95% COP Sway Area (mm^2)							
EMM + SE	402.22 ± 140.70	492.08 ± 140.70	471.18 ± 140.70	0.26	2, 24	0.776	0.02
[95% CI]	[110.40–694.04]	[200.26–783.91]	[179.36–763.00]				
n	13	13	13				
Sparta Balance Scan nonlinear metrics							
MVE							
EMM + SE	0.98 ± 0.03	1.00 ± 0.03	0.97 ± 0.03	1.06	2, 36	0.358	0.06
[95% CI]	[0.91–1.04]	[0.94–1.07]	[0.91–1.04]				
n	19	19	19				

Table 2. *Cont.*

	Condition			ANOVA			
Variable	**Baseline**	**Post-Kicking**	**Post-Heading**	***F***	**df**	***p***	**η^2**
mMSE-XY							
EMM + SE	0.35 ± 0.02	0.34 ± 0.02	0.33 ± 0.02	0.83	2, 36	0.446	0.04
[95% CI]	[0.32–0.38]	[0.31–0.37]	[0.30–0.36]				
n	19	19	19				
mMSE-2							
EMM + SE	0.81 ± 0.07	0.71 ± 0.07	0.81 ± 0.07	0.64	2, 36	0.535	0.03
[95% CI]	[0.66–0.96]	[0.56–0.86]	[0.66 ± 0.96]				
n	19	19	19				
mMSE-XY-2							
EMM + SE	0.21 ± 0.02	0.23 ± 0.02	0.21 ± 0.02	0.94	2, 36	0.400	0.05
[95% CI]	[0.17–0.25]	[0.19–0.27]	[0.17–0.25]				
n	19	19	19				
Kicking Leg Balance							
EMM + SE	184 ± 7.52	197 ± 7.52	201 ± 7.52	2.51	2, 36	0.096	0.12
[95% CI]	[169–199]	[181–212]	[186–216]				
n	19	19	19				
Non-Kicking Leg Balance							
EMM + SE	198 ± 6.81	207 ± 6.81	205 ± 6.81	1.38	2, 36	0.265	0.07
[95% CI]	[184–212]	[193–221]	[191–218]				
N	19	19	19				
COPx Shift (m)							
EMM + SE	0.147 ± 0.006	0.151 ± 0.006	0.156 ± 0.006	4.37	2, 36	0.020 *	0.20
[95% CI]	[0.133–0.160]	[0.137 ± 0.164]	[0.142–0.169]				
n	19	19	19				
Corticospinal inhibition							
cSP duration (ms)							
EMM + SE	152.66 ± 6.52	156.09 ± 5.87	162.03 ± 6.14	3.55	2, 15.82	0.053	0.31
[95% CI]	[138.66–166.67]	[143.64–168.54]	[149.00–175.06]				
n	320	320	340				

* denotes Omnibus test significance ($p < 0.05$). Italicised text denotes assessment type as subheading with accompanying metrics listed underneath. Abbreviations: CI-ML, complexity index in the mediolateral axis; CI-AP, complexity index in the anteroposterior axis; RMS, root-mean square; COP, centre of pressure; MVE, multiscale vertical entropy; mMSE, multivariate multiscale entropy; -XY, in the XY plane; -2, in dual stance; cSP, cortical silent period.

Table 3. Pairwise differences across experimental conditions.

Variable	Condition A		Condition B	Mean Difference	95% CI	%Δ	Cohen's d_z [± 95% CI]	p
Dual-stance postural control nonlinear measures (n = 13)								
CI-ML	Baseline	-	Post-Heading	0.65	[−7.50, 8.80]	−2.06	0.04 [± 1.09]	0.877
	Baseline	-	Post-Kicking	2.20	[−5.95, 10.35]	−6.99	0.15 [± 1.09]	0.602
	Post-Kicking	-	Post-Heading	−1.55	[−9.7, 6.6]	5.29	0.10 [± 1.09]	0.713
CI-AP	Baseline	-	Post-Heading	4.75	[1.10, 8.40]	−18.6	0.71 [± 1.22]	0.018 *
	Baseline	-	Post-Kicking	1.20	[−2.45, 4.85]	−4.75	0.18 [± 1.10]	0.524
	Post-Kicking	-	Post-Heading	3.55	[−0.10, 7.20]	−14.55	0.53 [± 1.16]	0.069
Dual-stance postural control linear measures (n = 13)								
RMS-ML (mm)	Baseline	-	Post-Heading	0.14	[−1.68, 1.96]	−4.09	0.04 [± 1.09]	0.880
	Baseline	-	Post-Kicking	−0.75	[−2.57, 1.07]	21.93	0.22 [± 1.10]	0.425
	Post-Kicking	-	Post-Heading	0.89	[−0.93, 2.71]	−21.34	0.27 [± 1.11]	0.345
RMS-AP (mm)	Baseline	-	Post-Heading	−2.18	[−4.18, −0.18]	39.25	0.60 [± 1.18]	0.042 *
	Baseline	-	Post-Kicking	−0.06	[−2.06, 1.94]	1.08	0.02 [± 1.09]	0.953
	Post-Kicking	-	Post-Heading	−2.12	[−4.12, −0.12]	37.77	0.58 [± 1.17]	0.047 *
COP Path Length (mm)	Baseline	-	Post-Heading	31.14	[−104.88, 167.16]	−3.55	0.12 [± 1.09]	0.658
	Baseline	-	Post-Kicking	−9.82	[−145.84, 126.20]	1.12	0.04 [± 1.09]	0.889
	Post-Kicking	-	Post-Heading	40.96	[−95.06, 176.98]	−4.62	0.16 [± 1.09]	0.561
Mean COP Velocity (mm/s)	Baseline	-	Post-Heading	0.48	[−1.79, 2.75]	−3.35	0.12 [± 1.09]	0.681
	Baseline	-	Post-Kicking	−0.16	[−2.43, 2.11]	1.1	0.04 [± 1.09]	0.889
	Post-Kicking	-	Post-Heading	0.65	[−1.62, 2.92]	−4.4	0.16 [± 1.09]	0.583
95% COP Sway Area (mm^2)	Baseline	-	Post-Kicking	−68.96	[−326.11, 188.19]	17.14	0.15 [± 1.09]	0.604
	Post-Kicking	-	Post-Heading	−89.86	[−347.01, 167.29]	22.34	0.19 [± 1.10]	0.500
	Baseline	-	Post-Kicking	20.90	[−236.25, 278.05]	−4.25	0.04 [± 1.09]	0.875
Sparta Balance Scan nonlinear metrics (n = 19)								
MVE	Baseline	-	Post-Heading	0.00	[−0.04, 0.04]	−1.02	0.01 [± 0.90]	0.956
	Baseline	-	Post-Kicking	−0.03	[−0.07, 0.01]	2.04	0.28 [± 0.92]	0.226
	Post-Kicking	-	Post-Heading	0.030	[−0.01, 0.07]	−3.00	0.30 [± 0.92]	0.207
mMSE-XY	Baseline	-	Post-Heading	0.02	[−0.02, 0.06]	−5.71	0.29 [± 0.92]	0.211
	Baseline	-	Post-Kicking	0.01	[−0.03, 0.05]	−2.86	0.18 [± 0.91]	0.439
	Post-Kicking	-	Post-Heading	0.01	[−0.03, 0.05]	−2.94	0.11 [± 0.90]	0.626
mMSE-2	Baseline	-	Post-Heading	0.00	[−0.2, 0.2]	0.00	0.00 [± 0.90]	0.997
	Baseline	-	Post-Kicking	0.10	[−0.1, 0.3]	−12.35	0.22 [± 0.91]	0.336
	Post-Kicking	-	Post-Heading	−0.10	[−0.3, 0.1]	14.08	0.22 [± 0.91]	0.334
mMSE-XY-2	Baseline	-	Post-Heading	0.01	[−0.03, 0.05]	0.00	0.06 [± 0.90]	0.800
	Baseline	-	Post-Kicking	−0.02	[−0.06, 0.02]	9.52	0.24 [± 0.91]	0.306
	Post-Kicking	-	Post-Heading	0.03	[−0.01, 0.07]	−8.7	0.30 [± 0.92]	0.204
Sparta Balance Scan linear metrics (n = 19)								
Kicking Leg Balance	Baseline	-	Post-Heading	−17.05	[−32.51, −1.59]	9.24	0.50 [± 0.95]	0.037 *
	Baseline	-	Post-Kicking	−12.53	[−27.99, 2.93]	7.07	0.36 [± 0.93]	0.121
	Post-Kicking	-	Post-Heading	−4.53	[−19.99, 10.93]	2.03	0.13 [± 0.90]	0.570
Non-Kicking Leg Balance	Baseline	-	Post-Heading	−6.95	[−18.44, 4.54]	3.54	0.27 [± 0.92]	0.243
	Baseline	-	Post-Kicking	−9.37	[−20.86, 2.12]	4.55	0.37 [± 0.93]	0.119
	Post-Kicking	-	Post-Heading	2.42	[−9.07, 13.91]	−0.97	0.09 [± 0.90]	0.682
COPx Shift (m)	Baseline	-	Post-Heading	−0.009	[−0.015, −0.003]	6.12	0.68 [± 0.90]	0.006 *
	Baseline	-	Post-Kicking	−0.004	[−0.04, 0.010]	2.72	0.30 [± 0.91]	0.204
	Post-Kicking	-	Post-Heading	−0.005	[−0.005, 0.005]	3.31	0.38 [± 0.92]	0.107
Corticospinal inhibition (n = 16)								
cSP duration (ms)	Baseline	-	Post-Heading	−9.37	[−18.35, −0.39]	6.14	0.51 [± 0.23]	0.058
	Baseline	-	Post-Kicking	−3.43	[−10.23, 3.37]	2.25	0.25 [± 0.22]	0.338
	Post-Kicking	-	Post-Heading ^	−5.94	[−10.41, −1.47]	3.81	0.63 [± 0.23]	0.019 *

Note: Mean differences calculated as Condition A–Condition B. * signifies significantly different ($p < 0.05$) pairwise comparisons. ^ *n = 17*. Italicised text denotes assessment type as subheadings with accompanying metrics listed underneath. All 20 cSP duration trials were used for analysis. Abbreviations: cSP, cortical silent period; CI-ML, complexity index in the mediolateral axis; CI-AP, complexity index in the anteroposterior axis; RMS-ML, root-mean square in the mediolateral axis; RMS-AP, root-mean square in the anteroposterior axis; COP, centre of pressure; MVE, multiscale vertical entropy; mMSE, multivariate multiscale entropy; -XY, in the XY plane; -2, in dual stance; cSP, cortical silent period.

3.1. Dual-Stance Postural Control Measures

A main effect of condition was observed for CI-AP ($\eta^2 = 0.23$, $F(2, 24) = 3.52$, $p = 0.046$), with the omnibus test reaching significance. *Post hoc* comparisons showed moderate decreases in CI-AP at post-heading testing compared with both baseline ($d_z = 0.71$, $t = 2.55$, $p = 0.018$) and post-kicking ($d_z = 0.53$, $t = 1.9$, $p = 0.069$), and a negligible decrease post-kicking compared with baseline ($d_z = 0.18$, $t = 0.65$, $p = 0.524$).

The trend of reduced CI-AP over time can be observed in Figure 5.

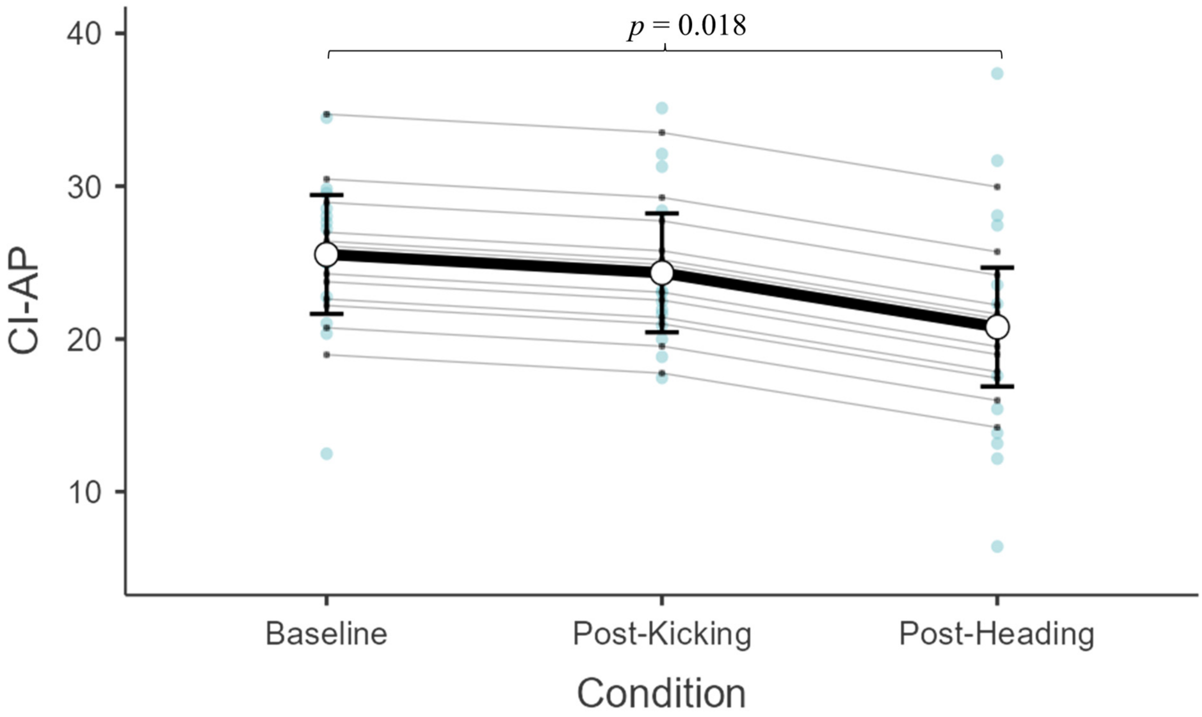

Figure 5. Within-participant differences in anteroposterior complexity index between conditions. White circles represent estimated marginal means (EMMs), blue dots indicate observed scores, black dots represent participant-level random effect estimates, and the connecting lines depict random slopes (grey = participant level; black = group level) across conditions. Error bars represent 95% confidence intervals of the EMMs. Abbreviations: CI-AP, complexity index in the anteroposterior axis.

A large effect of condition was observed for RMS-AP (η^2 = 0.20), though the omnibus test did not reach significance (F(2, 24) = 3.00, p = 0.069). Post hoc testing showed moderate (d_z = 0.60) RMS-AP increases at post-heading (t = −2.15, p = 0.042) and post-kicking conditions (d_z = 0.58, t = −2.09, p = 0.047) relative to baseline. Negligible effects were observed across CI-ML (η^2 = 0.01, F(2, 24) = 0.15, p = 0.863), RMS-ML (η^2 = 0.04, F(2, 24) = 0.54, p = 0.591), path length (η^2 = 0.02, F(2, 24) = 0.19, p = 0.828), mean COP velocity (η^2 = 0.01, F(2, 24) = 0.17, p = 0.846), and 95% COP sway area (η^2 = 0.02, F(2, 24) = 0.26, p = 0.776) (Appendix A, Figure A1a–f).

3.2. Sparta Balance Scan Metrics

Negligible to small changes were observed in mMSE-XY, MVE, mMSE-2 and mMSE-XY-2 between conditions (d_z = 0.00–0.30, F(2, 36), p = 0.358–0.535) (Appendix B, Figure A2a–d). A large main effect of condition was observed for COPx shift (η^2 = 0.20, F(2, 36) = 4.37, p = 0.020), with omnibus test significance. Pairwise comparisons showed a large increase at the post-heading condition (d_z = 0.68, t = −2.95, p = 0.006) compared to baseline. Small increases in COPx shift were also observed at post-heading compared to post-kicking (d_z = 0.38, t = −1.66, p = 0.107) and at post-kicking compared to baseline (d_z = 0.30, t = −1.29, p = 0.204). Kicking-leg balance moderately increased with statistical significance at post-heading relative to baseline (d_z = 0.50, t = −2.16, p = 0.037). A small increase in non-kicking leg balance was observed at post-kicking (d_z = 0.37, t = −1.60, p = 0.119) compared to baseline, though the difference at post-heading compared to post-kicking was negligible (d_z = 0.09, t = 0.41, p = 0.682). Between-condition trends in linear metrics of the Balance Scan can be viewed in Appendix C, Figure A3a–c.

3.3. Cortical Silent Period Results

A large effect of condition was observed for cSP duration ($\eta^2 = 0.31$, $F(2, 15.82) = 3.55$, $p = 0.053$), though the omnibus test did not reach conventional significance. However, a large estimated marginal mean (EMM) increase of 5.94 ms was observed at post-heading compared with post-kicking ($d_z = 0.63$, $t = 2.6$, $df = 15.82$, $p = 0.019$). Additionally, there was a small ($dz = 0.25$) EMM increase of 3.34 ms at post-kicking ($t = 0.99$, $df = 15.82$, $p = 0.338$) and a moderate ($d_z = 0.51$) 9.37 ms increase at post-heading ($t = 2.05$, $p = 0.058$) compared with baseline (Appendix D).

4. Discussion

This exploratory study sought to investigate whether the performances of ten AF-related kicking and ten heading trials would elicit detectable changes in measures of dual-stance and single-leg postural control, and corticospinal inhibition. Prior research has presented mixed findings regarding the effects of AF-related activities on measures of postural control and brain neurophysiology [5,8,9,35,47,49], so it has remained inconclusive whether AF heading independently contributes to acute neurocognitive and neurophysiological changes. This study limited heading exposure to ten trials, reflecting newly established training guidelines [45]. The mean peak linear (16.33 ± 2.49 g) and angular (1016 ± 296 rad/s^2) head accelerations in the current study are comparable to peak linear (≤16 g) and angular (1271 ± 602 rad/s^2; [47]) head accelerations reported in previous studies, which describe using similar ball delivery speeds [5,9,47,49,50]. Results of the current study suggested a large negative effect of condition on CI-AP, and large positive effects on RMS-AP, COPx shift and cSP. Among these, omnibus tests indicated statistically significant condition effects for CI-AP and COPx. Paired comparisons further suggested moderate changes in CI-AP, RMS-AP, kicking leg balance, COPx shift and cSP at post-heading compared with baseline, as well as from post-kicking to post-heading in CI-AP, RMS-AP, and cSP.

4.1. Dual-Stance Postural Control

A reduction in postural control complexity generally reflects impaired postural regulation, as seen in older adults [30,32] and in athletes following SRC [33,34]. Measures of postural control complexity have also repeatedly shown greater sensitivity than traditional linear metrics in detecting post-SRC impairments [19,29]. In the present study, a moderate reduction in CI-AP was observed at post-heading compared with baseline ($d_z = 0.71$). A further moderate reduction in CI-AP was evident between the post-kicking and post-heading timepoints, coinciding with moderate increases in RMS-AP observed at post-heading relative to both baseline and post-kicking. This pattern of reduced complexity with concurrent increases in sway amplitude is consistent with previous findings linking sway range with anteroposterior complexity following SRC [19]. Notably, complexity deficits in that study persisted for weeks beyond the resolution of sway amplitude abnormalities [19]. Considering those findings, the observed post-heading decline in CI-AP in the current study could suggest some reduction in postural control capacity following the collective performance of ten kicks and ten headers. This interpretation aligns with the loss of complexity hypothesis, which associates decreases in complexity with diminished adaptability [31]. Reductions in postural control complexity have been widely reported in athletes recovering from SRC [19,33,77], making it plausible that the performance of ten headers in the current study contributed to a detectable level of acute non-concussive injury, reflected by a measurable loss of postural control complexity. This perceived loss of complexity may correspond with subtle neurochemical and microstructural alterations, as reported in studies involving twenty [47] and forty [78] controlled AF headers. Furthermore, RMS-AP increases concurrent with CI-AP changes in this study may also reflect an acute response to the ten AF headers. However, previous work has found such changes to resolve more rapidly than deficits in complexity, underscoring the greater sensitivity of nonlinear measures [19].

Relatively small changes were observed in mediolateral complexity (CI-ML), consistent with similarly small changes in mediolateral sway amplitude (RMS-ML) across the session. In contrast, Haran et al. [8] reported larger increases in RMS-ML following ten headers, though this finding may have reflected their postural control assessment methods, which involved dynamic environmental and somatosensory perturbations. Such perturbations may have placed greater demands on the postural control system, contributing to larger sway amplitudes relative to baseline. However, because these perturbations require large, operationally cumbersome equipment, their transferability to field environments is limited [25]. Interestingly, Haran et al. [8] also reported larger increases in both ML and AP sway amplitude at 24 h after heading activity, compared with 1 and 48 h follow-ups, attributing this delayed impairment to the onset of glucose hypometabolism. This finding highlights the potential value of conducting postural control assessments over longer time intervals to examine interactions between head-impact-related biochemical changes and motor outcomes. Future studies incorporating postural control complexity measures across longer follow-up intervals may be particularly informative in reporting post-NCHI responses, given the demonstrated sensitivity of complexity measures to subtle deficits following SRC [19].

The current study is the first to investigate the acute effects of NCHIs—specifically via AF heading—on postural control complexity using MSE. The reduction in CI-AP observed at post-heading relative to post-kicking and baseline may reflect a reduced adaptability in the postural control system following AF heading, highlighting the need for further investigation. Nonlinear postural control measures have potential as non-invasive, objective indicators of acute movement deficits following exposure to NCHIs. The use of portable force platforms could enhance the feasibility of conducting such assessments in field environments, providing an accessible approach for objectively monitoring postural control adaptations in athletes.

4.2. Sparta Balance Scan Performance

The Sparta Balance Scan is a commercially available balance assessment tool, designed to quantify postural control using a portable force platform suitable for field-based assessment. In this study, the Sparta system was included to evaluate its potential application in detecting acute changes in postural control (i.e., balance) following exposure to NCHIs. Accounting for field transferability, the Sparta Balance Scan utilises relatively short stance target windows. While these target windows are convenient for field use, it is important to note that they are considerably shorter than typical single-leg trial durations of $\geq$20 s [79–81], and recommended dual-stance trial durations of $\geq$60 s [82,83], which enable sufficient capture of COP oscillations as low as 0.01 Hz. Consequently, the brevity of the Balance Scan target windows should be acknowledged as a methodological consideration when interpreting its outcomes.

While previous research has shown evidence of single-leg balance deficits in athletes with SRC history and individuals with brainstem lesions [84,85], the effects of repetitive NCHI exposure on single-leg balance remain under-investigated. In the current study, small-to-moderate improvements in kicking and non-kicking leg balance performance were observed at post-kicking and post-heading relative to baseline; however, differences at post-heading relative to post-kicking were negligible. The pattern of improved single-leg balance performance relative to baseline may reflect a within-session learned effect by the

AF-trained participants, particularly as prior evidence has demonstrated a positive correlation between higher AF-competition level and superior unipedal postural performance [86]. Although competition level was not a factor in recruitment for this study, the prerequisite of at least two years of previous AF training likely suggests some familiarity with single-leg stance control (i.e., via kicking performance), which may have contributed to the apparent within-session learning effects observed relative to baseline.

Small-to-moderate increases in COPx shift were also observed between conditions, including a moderate increase reported at post-heading relative to baseline. Consistent with the pattern of single-leg balance scores, the change in COPx shift may also reflect increased familiarity with the Balance Scan protocol across the session. Previous research has compared the effects of dual-to-single-leg stance transitions with and without visual feedback, finding that despite COP x- and y-axis amplitudes remaining relatively unchanged, healthy participants demonstrate accelerated COPx shifting under visual feedback conditions [87]. This finding was interpreted as a reflection of enhanced predictive motor processing capacity when using visual feedback, accelerating the transitions from dual- to single-leg stance [87]. While evidence regarding post-concussive changes in COPx excursions during stance transitions is limited; research in gait initiation (transitioning from standing posture to cyclic gait) has shown that participants recovering from SRC exhibit smaller COPx excursions than healthy controls, interpreted as impaired preparatory postural adjustments [88]. Collectively, this evidence suggests that the observed COPx shift changes were unlikely to result from exposure to ten NCHIs via AF heading. It is perhaps more likely that participants grew familiar with the protocol of the Balance Scan, particularly accustoming themselves to its live COP trace feedback. As discussed with the linear measures of dual-stance postural control, the single-leg balance and mediolateral shift metrics of the Sparta Balance Scan may lack the required sensitivity to detect subtle structural changes in balance dynamics. Future research should further evaluate the sensitivity of these metrics to within-session learning effects, which may have influenced the current pattern of results.

In addition to linear postural control metrics, the Sparta Balance Scan also quantifies postural complexity using multiscale vertical entropy (MVE) and multivariate multiscale entropy (mMSE) metrics. MVE reflected vertical force complexity across single-leg stance phases, though small changes between conditions were minimal and of uncertain physiological relevance. Alternatively, though less established than univariate MSE, mMSE has been described as a consistent, reliable parameter in detecting between-condition differences [72]. By jointly analysing multiple signal axes (anteroposterior and mediolateral, including the vertical axis in mMSE-2) rather than treating each in isolation, mMSE is designed to capture inter-signal axis interactions and provide a more comprehensive assessment of postural control dynamics [89]. However, as no previous study has reported mMSE values derived from the Sparta Balance Scan, its relative novelty and limited uptake warrant some caution when interpreting results within the broader postural control literature. A small, non-significant reduction in mMSE-XY was observed at post-heading relative to baseline in the current study (Appendix B, Figure A2b), possibly reflecting a reduction in single-leg postural control adaptability across the session. The decrease in mMSE-XY is consistent with the reduction in CI-AP determined from the dual-stance postural control assessment, albeit at a much smaller scale. Additionally, only negligible-to-small changes were observed in mMSE-2 and mMSE-XY-2 between conditions. Given that previous research has recommended trial periods of beyond 60 s to appropriately capture changing postural dynamics [82,83], it is possible that the Balance Scan's target windows of four five-second single-leg stance phases and one two-second phase of dual stance at the beginning of the

Scan were too short to capture meaningful changes in postural complexity using these metrics.

Despite limited reporting of the Sparta Balance Scan metrics in peer-reviewed literature, existing data suggest the between-session reliability of dominant and non-dominant single-leg balance scores ranges from poor to good (coefficient of variation [CV] = 4–6%) [90]. Given the system's relatively unexplored postural control metrics in published research, further validation is warranted, particularly in the assessment of post-NCHI responses. Future research should confirm the reliability and sensitivity of the Sparta Balance Scan's postural control metrics, thereby informing its refinement as an accessible and valid field-based tool for assessing postural control, including complexity. The portability and user-friendly design of the Sparta Balance system offer promise as an objective alternative in the field-based assessment of postural control.

4.3. Cortical Silent Period

As a marker of corticospinal inhibition, cSP is commonly reported in the assessment of acute neurophysiological impairment following mTBI or exposure to repetitive head impacts [5,49,54,55]. Increases in cSP following head impact exposure are believed to represent increased gamma-amino-butyric acid (GABA)ergic-mediated inhibition in the corticospinal tract [49]. In the current study, the estimated mean cSP lengthening of 5.9 ms after ten headers is comparable to statistically significant increases of 5.4 ms and 6.8 ms previously reported following twenty headers [5,49]. Of all metrics reported in this study, cSP exhibited the largest eta-squared effect of condition, despite a non-significant omnibus p value. Pairwise comparisons showed moderate cSP increases at post-heading, relative to both baseline and post-kicking, with the increase from post-kicking to post-heading reaching statistical significance. These findings tentatively suggest an effect of heading on corticospinal inhibition, aligning with a growing body of evidence in this area [5,49]. The results of the current study therefore imply that even a relatively small number of headers (ten in the current study) could induce changes in cSP. However, further confirmatory research with a larger sample is required to confirm this speculation.

Although Hamel et al. [49] reported a significant increase in cSP following twenty headers, a comparable lengthening of 6.7 ms was also observed following a control AF task, with no between-group differences detected. The authors acknowledged that including both practised and unpractised participants in the heading group may have influenced these outcomes, as neck musculature activation and heading technique can moderate head acceleration and vary considerably across skill levels [91]. Such factors may account for the inter-individual variability observed in both Hamel's and the present study's cSP responses (Appendix D) and may also influence postural control outcomes. Future research should therefore incorporate concurrent measurement of neck muscle activity and heading technique to better contextualise between-participant differences in post-heading responses.

While mTBIs are known to influence inhibitory [55,92] and motor control [93,94] processes, the impact of NCHIs on these outcomes remains unclear. Prior work has linked SRC with an increased risk of non-contact musculoskeletal injury, particularly through altered knee kinematics during cutting [95] and landing tasks [96]. Therefore, understanding whether NCHIs may also elicit inhibitory effects is of both practical and clinical importance. The present findings showed a moderate increase in cSP and a concurrent, statistically significant reduction in CI-AP post-heading. Although this study did not directly assess associations between these outcomes, similar patterns of increased corticospinal inhibition and altered motor function have been described elsewhere, with between-group differences in inhibition linked to differences in mobility [97] and motor coordination [98]. Swanson et al. [98] observed that greater corticospinal inhibition in older adults was associated with enhanced gait coordination, suggesting that inhibitory changes do not always imply dysfunction and may, in some contexts, reflect compensatory adaptation. Nonetheless, reductions in mobility and motor coordination have been associated with impaired daily motor function [97] and an elevated risk of non-contact musculoskeletal injury [99]. Although correlations between corticospinal inhibition and postural complexity were not examined in this study, future work should investigate whether acute postural control adaptations statistically correlate with changes in corticospinal inhibition following exposure to NCHIs.

4.4. Limitations

The prescription of ten headers in this study adheres to heading guidelines set by the English Football Association. Consequently, the magnitude of acute post-heading responses may have been reduced compared with previous studies utilising 20 headers [5,49]. Additionally, this study controlled for heading technique (linear headers), rather than instructing participants to head the ball to either side. Evidence suggests that head impacts involving rotational motion may produce a more injurious mechanism than linear motion, generating greater shear forces on the brain [100,101]. Therefore, inclusion of rotational headers in the current study could potentially have yielded larger post-heading effects. However, linear headers are reportedly more pervasive than rotational headers during normal match play [102].

The heading protocol of ten headers in ten minutes represents a greater frequency than typically observed during competition. Prior research has reported a mean match exposure of 4.2 headers per player at an elite men's level, with a maximum of 10.6 headers across 90 min [102]. Future research could therefore examine acute responses to heading over a session duration that more closely resembles normal match play (i.e., 1.5–2 h). However, the current protocol was intentionally designed to replicate a heading-focused training drill, where heading frequency may exceed that seen in competition, while remaining within the ten-header limit recommended by the English Football Association's Heading Guidance [45].

The use of a circular TMS coil may have limited the locality and efficacy of the electromagnetic stimuli. If available, a figure-eight coil would have been preferred to provide a more focused stimulation at the MEP site, perhaps resulting in more reliable cSP measurements [103,104]. Furthermore, unforeseen data capture issues identified during analysis reduced the completeness of some measurements, resulting in lower statistical power for certain comparisons. These factors may have influenced the ability to detect significant effects.

Despite attempts to recruit both male and female cohorts for participation in this study, the recruited sample was limited to adult males aged 18–50 years. As such, the generalisability of the results is limited to this demographic. Importantly, previous research has demonstrated greater peak linear and angular head accelerations during controlled AF heading [105] and longer SRC recovery timescales [106] among female athletes relative to males. Additionally, head and neck size may be predictive of linear and angular accelerations during AF heading [107], though age-related effects are uncertain [105]. Therefore, future studies should target female and youth-aged AF players to determine if acute responses to ten headers differ based on sex and age. In addition, the limited participant sample reduced the experimental protocol to a single order of kicking and then heading, without the ability to rearrange the order of our intervention activities. Consequently, despite attempts to match activity levels and rest periods, it is uncertain if the results observed post-heading were not cumulatively affected by confounding factors such as activity-related excitability or a degree of fatigue onset. However, as heading normally occurs within a broader context of training or match play, some degree of prior physical activity is expected. This could therefore enhance the representativeness of our findings in this respect.

5. Conclusions

This exploratory study investigated the acute effects of AF-related kicking and heading activity on linear and nonlinear measures of postural control and corticospinal inhibition. The main finding of this study was a large, negative effect of condition on CI-AP, with pairwise differences showing moderate decreases in CI-AP at post-heading relative to post-kicking and baseline, but not post-kicking relative to baseline. This suggests a deficit in postural control complexity following the performance of ten headers, possibly reflecting a transient loss of adaptive capacity [31,108]. This apparent deficit in postural complexity is consistent with changes related to mTBI [19,33,34]; however, this is the first study to suggest this deficit following NCHI exposure. In parallel, moderate increases in RMS-AP, kicking leg balance and COPx were observed across the protocol, perhaps suggesting some utility of these metrics in detecting change. Additionally, moderate increases in cSP duration observed at post-heading relative to both baseline and post-kicking suggest a possible acute neurophysiological response to ten headers. This is consistent with previous evidence demonstrating cSP lengthening immediately following twenty headers [5,49].

The current exploratory study demonstrates a potential sensitivity of anteroposterior postural complexity and cortical silent period as respective measures of acute postural dynamic and corticomotor adaptations following exposure to ten controlled football headers and ten kicks. Future confirmatory research should investigate the effects observed in the current study with a larger sample and further evaluate sensitive, field-based approaches for assessing responses to non-concussive head impacts.

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Conflicts of Interest: A.J.P. is currently a nonexecutive director of the Concussion Legacy Foundation Australia. He is remunerated for expert advice to medico-legal practices. The authors declare no other conflicts of interest. Although there was some collaboration with Sparta Science in the provision of their force platform system and consultation regarding experimental practices, the company was not involved in the analysis or reporting of data in this manuscript.

Appendix A. Within-Participant Mean Differences in Dual-Stance Postural Control Measures Between Conditions

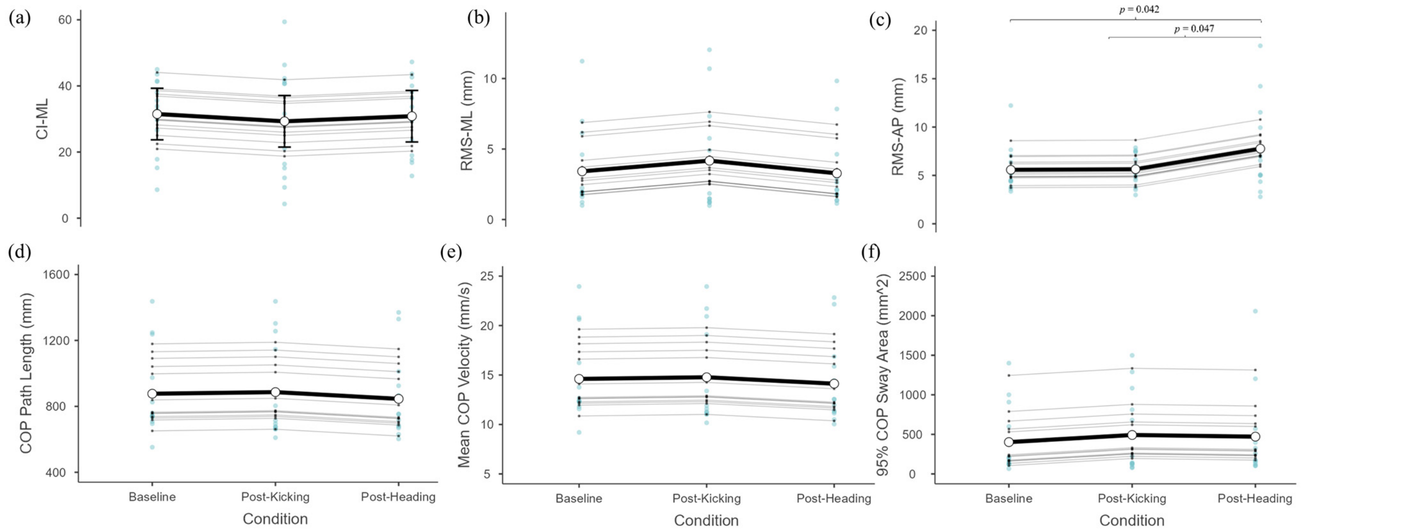

Figure A1. Conditions include Baseline, which involves initial testing at beginning of session; Post-Kicking, which involves re-testing following the performance of ten kicking trials; and Post-Heading, which involves re-testing following the performance of ten heading trials. White circles represent estimated marginal means (EMMs), blue dots indicate observed scores, black dots represent participant-level random effect estimates, and the connecting lines depict random slopes (grey = participant level; black = group level) across conditions. Error bars represent 95% confidence intervals of the EMMs. Brackets with denoted *p*-values denote significant estimated marginal mean differences between conditions. Subfigures present between-condition EMM trends in: (**a**) complexity index in the mediolateral axis, (**b**) root mean square of the instantaneous centre of pressure (COP) displacements from the mean position in the mediolateral axis, (**c**) root mean square of the instantaneous COP displacements from the mean position in the anteroposterior axis, (**d**) total centre of pressure path length, (**e**) mean COP velocity, (**f**) 95% confidence ellipse area of the total COP sway. Brackets with denoted *p*-values denote significant estimated marginal mean differences between conditions. Abbreviations: CI complexity index; -ML, in the mediolateral axis; RMS, root-mean square; -AP, in the anteroposterior axis; COP, centre of pressure.

Appendix B. Within-Participant Mean Differences in Sparta Balance Scan Nonlinear Metrics Between Conditions

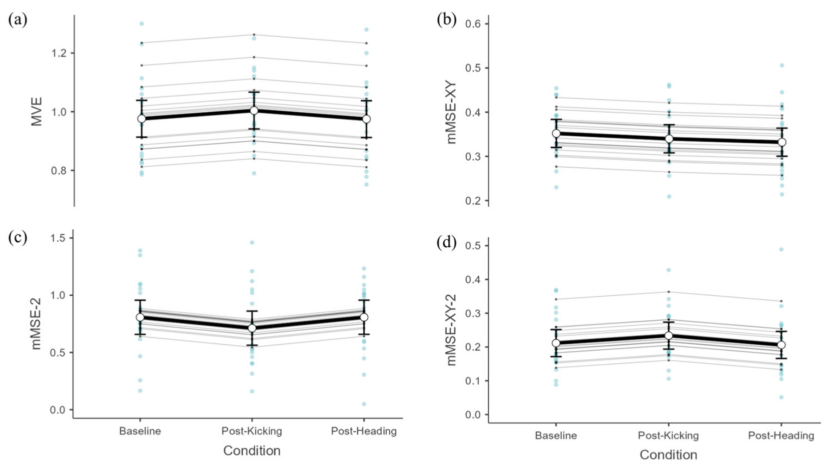

Figure A2. Conditions include Baseline, involving initial testing at beginning of session; Post-Kicking, involving re-testing following the performance of ten kicking trials; and Post-Heading, involving re-testing following the performance of ten heading trials. White circles represent estimated marginal means (EMMs), blue dots indicate observed scores, black dots represent participant-level random effect estimates, and the connecting lines depict random slopes (grey = participant level; black = group level) across conditions. Error bars represent 95% confidence intervals of the EMMs. Subfigures present between-condition EMM trends in: (**a**) multiscale vertical entropy, (**b**) multivariate multiscale sample entropy in the XY plane, (**c**) multivariate multiscale entropy in dual stance, (**d**) multivariate multiscale entropy in the XY plane in dual stance. Abbreviations: MVE, multiscale vertical entropy; mMSE, multivariate multiscale entropy; -XY, in the XY plane; -2, in dual stance.

Appendix C. Within-Participant Mean Differences in Sparta Balance Scan Linear Metrics Between Conditions

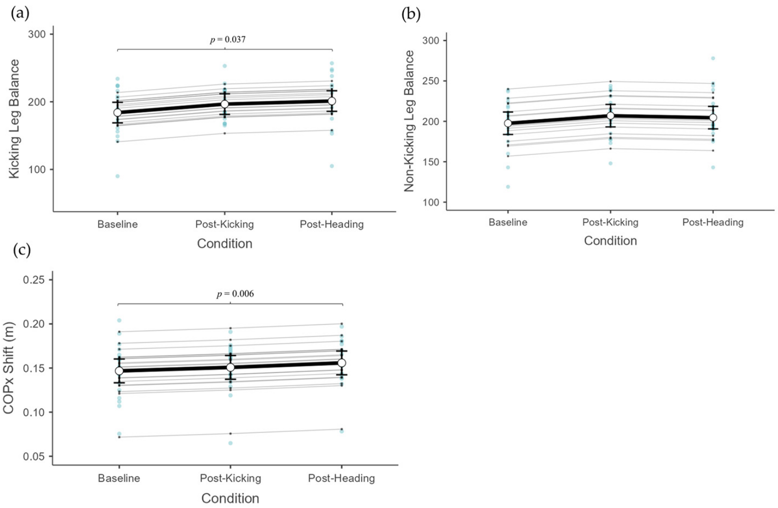

Figure A3. Conditions include Baseline, involving initial testing at beginning of session; Post-Kicking, involving re-testing following the performance of ten kicking trials; and Post-Heading, involving re-testing following the performance of ten heading trials. White circles represent estimated marginal means (EMMs), blue dots indicate observed scores, black dots represent participant-level random effect estimates, and the connecting lines depict random slopes (grey = participant level; black = group level) across conditions. Error bars represent 95% confidence intervals of the EMMs. Subfigures present between-condition EMM trends in: (**a**) kicking leg balance, (**b**) non-kicking leg balance, and (**c**) centre of pressure shift in the mediolateral axis. Brackets with denoted *p*-values denote significant estimated marginal mean differences between conditions. Abbreviations: COPx, centre of pressure in the mediolateral axis.

Appendix D. Within-Participant Mean Differences in Cortical Silent Period Duration, Representing Corticospinal Inhibition

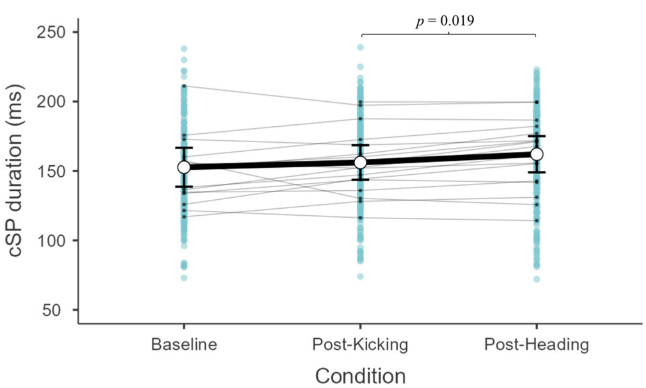

Figure A4. Conditions include Baseline, involving initial testing at beginning of session; Post-Kicking, involving re-testing following the performance of ten kicking trials; and Post-Heading, involving re-testing following the performance of ten heading trials. White circles represent estimated marginal means, blue dots indicate observed scores, black dots represent participant-level random effect estimates, and the connecting lines depict random slopes (grey = participant level; black = group level) across conditions. Error bars represent 95% confidence intervals of the EMMs. A significant estimated marginal mean difference is denoted between post-kicking and post-heading conditions. Abbreviation: cSP, cortical silent period.

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