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The use of continuous spectral analysis for the assessment of postural stability changes after sports-related concussion

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ABSTRACT

Impaired postural stability is associated with a variety of pathologies including sports-related concussion (SRC). Quantification of centre of pressure (COP) movement is the most common focus of instrumented assessment. Frequency-domain COP analyses have focused primarily on summary measures or pre-defined frequency bands but continuous analysis may provide novel and complementary insight into pathological control mechanisms. Our aims were (i) to compare post-SRC COP trajectory changes identified using clinician scores (Modified Balance Error Scoring System (M-BESS)), time-domain COP variables and continuous frequency spectral comparison; and (ii) to characterise frequency spectra changes. Male rugby players aged 15–19 years (n = 135) completed a pre-season baseline assessment comprising vision-obscured double-leg, single-leg and tandem stances on a force platform. Participants diagnosed with SRC during the season (n = 15) underwent repeat testing (median 4 days post-SRC; IQR 2.5–6.5). Baseline and post-SRC COP trajectories were compared using common time-domain COP variables, M-BESS scores and continuous frequency spectra. Post-SRC changes were identified using all three approaches. Spectral analysis revealed the largest effect size (Cliff's delta 0.39) and was the only method to identify differences in all three stances and in double-leg stance. All post-SRC increases in spectral content were in the anteroposterior direction; all decreases were in the mediolateral direction. Changes were localised to higher frequencies (1.7–8 Hz) except for double-leg stance anteroposterior direction, for which increases were observed throughout the analysed range. Our findings suggest that this method of spectral comparison may provide a more responsive and meaningful measure of postural stability changes after SRC than other commonly-used variables.

1. Introduction

The control of postural stability involves integrated contributions from visual, vestibular and proprioceptive systems to maintain balance (Horak and Macpherson, 1996). Impairments are associated with a wide range of pathologies including neurological disorders and musculoskeletal injuries. Visual balance scoring systems and instrumented assessment are both used clinically for evaluating postural stability: the former have the advantage of needing no technical equipment so are widely used in clinical environments and the latter offers improved objectivity and sensitivity (Horak, 1997; Ruhe et al., 2014; Visser et al., 2008). The most common variables utilised in instrumented assessment are summary measures describing movement of the centre of pressure (COP) in the time domain (Crétual, 2015; Paillard and Noé, 2015; Palmieri et al., 2002), although evidence for their reliability and validity for concussion diagnosis and monitoring of recovery is mixed (Murray et al., 2014).

Sport-related concussion (SRC) is a complex brain injury resulting from mechanical trauma and associated with a range of neurological impairments including effects on motor control (McCrory et al., 2017, 2001). Many athletes with SRC demonstrate altered balance (Baracks et al., 2018; Guskiewicz, 2011; Guskiewicz et al., 2001; Howell et al., 2019; Parrington et al., 2019; Powers

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et al., 2014; Valovich McLeod and Hale, 2015) and the Modified Balance Error Scoring System (M-BESS) protocol for clinical balance evaluation (McCrory et al., 2017; Riemann and Guskiewicz, 2000) is an integral part of the Sport Concussion Assessment Tool (SCAT, 2013) used to assist with acute assessment. Instrumented assessment of SRC initially focused on time-domain COP variables (Goldie et al., 1989; Palmieri et al., 2002; Riemann et al., 1999), with more-recent studies investigating non-linear measures of COP movement to quantify properties such as entropy and complexity (Cavanaugh et al., 2005; Haid and Federolf, 2018; Sosnoff et al., 2011) and also exploring the use of inertial sensor instead of force platform technology to characterise body sway (Doherty et al., 2017; King et al., 2014). The movement of the COP can be analysed in the frequency domain as well as the time domain, and frequency domain analyses have been employed for balance assessment of patients with a variety of pathologies (Degani et al., 2017; Golomer et al., 1994). Limited analysis in this domain has previously been reported for the assessment of postural stability changes post-SRC, but participants with a history of mild traumatic brain injury have been found to have a lower frequency containing 80% of spectral power than controls in bipedal stance with feet parallel (Degani et al., 2017).

The multiple postural control systems involved in the regulation of balance have different time delays in their control pathways, enabling their relative afferent contributions to be studied by identifying characteristic COP frequency responses.

The weighting of these system contributions to postural control can be modulated by pathology, by the availability of information through each modality and by the perceived reliability of the information (Barela et al., 2014; Creath et al., 2008; Hwang et al., 2016; Jeka et al., 2000; Peterka, 2002; Polastri et al., 2012). The vestibular and visual systems both appear to function primarily at frequencies below 0.2–0.5 Hz, the cerebellar system at approximately 0.5–2 Hz and proprioceptive systems at >2 Hz (Diener et al., 1984; Fitzpatrick et al., 1992; Mauritz and Dietz, 1980; Nashner, 1976; Paillard and Noé, 2015), although there is a lack of consensus regarding the precise range of each band (Kanekar et al., 2014; Palmieri et al., 2002). Analysis of the COP time series in the frequency domain rather than the time domain permits the structure of the COP trajectory to be quantified and related to contributions of these different systems, providing a basis for insight into the underlying impairment (Palmieri et al., 2002; Rougier, 2008).

Frequency domain analysis has most-commonly focused on summary measures of frequency content such as average frequency or frequency below which 80% of spectral power is contained. When comparisons across the full frequency range are undertaken, signal content is typically divided into three frequency bands corresponding to the control systems described above, and the energy content in each band then compared between groups (although higher-resolution approaches have also been utilised (Singh et al., 2012; Soames and Atha, 1982)). There are several key limitations of this approach. Firstly, *a priori* designation of frequency band boundaries is required so arbitrary selection differences can modify findings. Secondly, the sensitivity of this method to alterations in narrower and/or non-predefined frequency band content is limited by low-resolution comparisons. For example, the ability to detect change in the ankle flexor myotatic reflex response, reported to be localised to 4–5 Hz (Dietz et al., 1983), can be lost when only overall change in a broader-range 'high frequency' band is analysed. An alternative method to overcome these limitations is to analyse the spectra continuously across a complete frequency range. This approach has previously been utilised in the field of neurophysiology (Amjad et al., 1997; Diggle, 1990; Halliday et al., 1995; Halliday and Rosenberg, 1999) but has yet to be applied to COP data for the investigation of pathological changes. The technique has the potential to provide novel insight into neural control mechanisms associated with pathological postural stability and to reveal clinically meaningful effects that can be used to probe control system impairments without the need to predefine frequency band boundaries.

Here we demonstrate the potential utility of this method for analysis of COP time series data to evaluate spectral content changes following SRC. Our first aim was to investigate whether the spectral analysis technique was more responsive than standard COP variables and clinician scores in identifying post-concussive balance deficits. We compared the differences and effect sizes identified in double leg, single leg and tandem stance positions using the M-BESS clinician scoring system, common time-domain COP variables and continuous frequency spectra pre-SRC with those measured in the same participants post-SRC. We hypothesised that the spectral analysis technique would identify differences in more stance-direction combinations and with larger effect sizes than the other two methods, as the structure of the whole signal is compared. We expected greater differences for all methods to be observed in the single-leg and tandem stance positions than in the double leg stance due to the increased challenge to postural stability.

In addition to enhanced ability to identify post-SRC changes, a potential benefit of frequency-domain analysis is that changes in neuromuscular control strategies can be investigated to aid understanding of how the deficit may have arisen. The second aim of the study was thus to describe and characterise the continuous frequency spectral changes after SRC. Based on previous findings, we hypothesised that we would find differences commensurate with a shift towards lower frequency components in all stance positions and in both the anteroposterior and mediolateral directions.

2. Methods

The investigation was conducted as part of a larger study on baseline and post-SRC outcome measures in schoolboy rugby players (Cosgrave et al., 2018). All eligible students (aged 15–19 years and no pre-existing neurological disorders or history of neurological disorders) on the rugby union teams of five secondary schools in Dublin, Ireland (n = 211) were invited to participate. Ethical approval was provided by the Sports Surgery Clinic Research Ethics Committee. All participants and their parent/guardian provided informed written consent.

All participants who chose to enroll in the study (n = 135) attended a pre-season baseline session in which they completed a battery of screening tests. Any participant who was diagnosed with a SRC during the following rugby season repeated the testing battery as soon as possible after the injury, and weekly thereafter until fully recovered. Median time from injury to first post-SRC test was 4 days (IQR 2.5–6.5). SRC was diagnosed by a medical professional independent of the study according to best guidelines for clinical practice and the diagnosis confirmed on presentation at the study centre by a consultant Sports and Exercise Medicine physician using the SCAT3 questionnaire, neurological screening and a full clinical assessment. Sixteen participants were diagnosed with SRC during the study period. Of these, one was excluded due to technical issues with data collection. Balance assessment data from the baseline screening session and the first post-injury assessment of the remaining 15 were analysed. Five of these participants reported that they had obtained at least one previous SRC; all reported that they were no longer affected by the injury. None had previously been diagnosed with ADHD or any hearing/auditory processing disorder; one had had a previous knee arthroscopy. At the time of the post-SRC session, 13 of the 15 participants were still experiencing symptoms as assessed using the SCAT3

Symptom Evaluation questionnaire. Of these, median number of symptoms reported was 9.5 and median total symptom severity score was 14; three reported balance problems. Two participants were diagnosed with two or more SRCs during the study: only the post-SRC dataset collected after the first SRC (and before the second SRC) was analysed.

The balance assessment comprised a modified Balance Error Scoring System (M-BESS) protocol as per SCAT3 (2013). Participants completed three stance tasks on a multi-axis force platform (1000 Hz; BP400600, AMTI, USA): (i) double-leg stance (DL) with the feet placed side by side as close to each other as possible; (ii) single-leg stance (SL) on the non-dominant leg; (iii) tandem stance (TAN) with the feet placed heel-to-toe and the non-dominant foot located most posteriorly (Fig. 1). Resting the dominant leg on the non-dominant leg in SL was not permitted. All stances were performed barefoot with eyes closed and were held for 20 s, after an initial three-second initiation phase in which the start position was established. An observer recorded the error score for each stance as per SCAT3: types of errors included stepping or falling, opening the eyes, and lifting the hands off the iliac crest (SCAT3, 2013).

COP position data were filtered using a 4th order zero-lag Butterworth filter with a corner frequency of 10 Hz (Ruhe and Walker, 2010). A *touchdown* was defined as the foot leaving the force platform or the opposite foot touching the ground, and the number of these was recorded for each trial. Data during the three-second initiation phase, the termination phase (after completion of the 20 s stance duration) and after the last COP change of direction before a touchdown (identified as an inflection point in either the AP or ML time series) were discarded prior to further analysis. For example, if a touchdown occurred 18.5 s, with the COP change of direction immediately before the touchdown occurring at time 18.2 s, the COP time series was truncated at 18.2 s before further analysis in order to avoid spectral artefacts resulting from interpolation. Touchdowns occurred in a total of 8 trials for the SL stance condition (4 at baseline and 4 post-SRC) and 6 trials for the TAN stance condition (3 at baseline and 3 post-SRC). No more than one touchdown occurred for any trial.

The discrete Fourier transform (DFT) of COP position was taken for each dataset in both the anteroposterior (AP) and mediolateral (ML) directions for each of the three stances at baseline and post-SRC to estimate frequency domain auto-spectra using a fast Fourier transform. Before computing the DFT a linear de-trend was performed and the data weighted by a Hanning window function to remove discontinuities at the start and end of the signal whilst minimising spectral leakage. Frequency resolution was 0.06 Hz. Pooled spectral estimates across participants were calculated (Amjad et al., 1997) and smoothed using a discrete spectral average of order 20 (Diggle, 1990). Analysis of the time series suggested that the assumption of stationarity was not unreasonable for our dataset (see 'Supplementary Stationarity Investigation' for further details).

Fig. 1. Foot placement for stance tasks. Double-leg stance (A), single-leg stance (B) and tandem stance (C).

Three analysis approaches were implemented: one using clinician scores, one using common discrete point variables extracted from the COP trajectories and one using the frequency domain spectra of the COP trajectories. The alternative hypothesis that baseline values differed from post-SRC values was tested for each stance and analysed parameter within the group who were diagnosed with an SRC. Significance was accepted at $\alpha = 0.05$.

Cliff's delta (Cliff, 1993) was calculated as a measure of standardised effect size for all significant differences. This statistic requires no assumptions regarding the shape of the underlying distribution and has been shown to be robust for small sample sizes with non-normal distributions, whilst also performing well for normally distributed continuous data (Delaney and Vargha, 2002; Vargha and Delaney, 2000). The absolute value of Cliff's delta (δ) ranges from 0 (no effect) to 1 (maximal effect). Thresholds for small, medium and large effect sizes of 0.11, 0.28 and 0.43 respectively have previously been calculated (Vargha and Delaney, 2000) based on the thresholds used for Cohen's d (Cohen, 1977). These may be used as a guide for interpreting δ.

2.1. Clinician scores analysis

A Wilcoxon's signed rank test was used for each stance, and for the total aggregate score across all stances, to test the alternative hypotheses that baseline M-BESS scores differed from post-SRC M-BESS scores.

2.2. Discrete point analysis

Six discrete point variables characterising the movement of the COP were calculated for each stance: maximum mediolateral ($range_{ML}$) and anteroposterior ($range_{AP}$) displacement range of the COP; standard deviation of mediolateral (SD_{ML}) and anteroposterior (SD_{AP}) position of the COP; mean velocity of the COP ($COPV_{mean}$); and 95% confidence ellipse area of the COP (CEA95%) (Prieto et al., 1996). These variables were selected because of their widespread use in the existing literature and because they characterise several different descriptors of the COP (Paillard and Noé, 2015; Palmieri et al., 2002). The number of touchdowns was also recorded and analysed. A Wilcoxon's signed rank test was used for each stance to test the alternative hypotheses that baseline results differed from post-SRC results.

2.3. Spectral analysis

A log ratio test, calculated using a log ratio of two spectral estimates, was used to compare baseline and post-SRC COP spectra for each of the three stance positions in the AP and ML directions (Diggle, 1990). The ratio is calculated as

$$log_{10}(sc_{POST} - sc_{PRE})$$

where sc_{POST} and sc_{PRE} are the pooled spectral coefficients for post-SRC and baseline data respectively. The null value based on the hypothesis of equal spectra is zero. Alpha was accepted at 0.05 and 95% confidence intervals were set using an F-distribution, with the null hypothesis rejected if these limits were exceeded by the variate (Diggle, 1990; Halliday et al., 1995). Frequency comparisons were reported in the range 0.1–10 Hz. All spectral analysis routines were implemented using the Neurospec toolbox for MATLAB (version 2.0, www.neurospec.org). Pointwise absolute δ was calculated across the frequency spectrum. Where frequency regions containing significant differences in spectral content were identified, mean absolute δ within each region was calculated and reported as a summary statistic.

3. Results

3.1. Clinician scores

Results are shown in Table 1. M-BESS scores for TAN stance were greater post-SRC than at baseline with a small effect size ($p = 0.05$, $W = 6$, $\delta = 0.19$). No significant differences in M-BESS scores were identified for DL, SL or total aggregate score.

3.2. Discrete point analysis

Results are shown in Table 2. $Range_{AP}$ for SL stance was smaller post-SRC than at baseline with a medium effect size ($p = 0.02$, $w = 89$, $\delta = 0.30$). No other differences were identified for any other variable in SL stance and no differences were identified for any variable in DL and TAN stance.

3.3. Spectral analysis

Results are shown in Fig. 2 and summarised in Table 3. Frequency regions of significant changes in signal content between baseline and post-SRC were identified for all stances. Post-SRC, a decrease in ML spectral content was identified in SL and TAN and an increase in AP spectral content was identified in DL and TAN. The DL increase was across the full analysed frequency range (0.1–10 Hz) and had the largest mean effect size identified in any analysis ($\delta = 0.39$). Changes in SL and TAN were within the range 1.7–8.0 Hz.

4. Discussion

This is, to the authors' knowledge, the first continuous high-resolution analysis of spectral frequency differences post-SRC. Post-SRC changes were identified using M-BESS, discrete point and spectral analysis approaches; spectral analysis was the only approach to identify differences in all three stances, the only method to identify differences in DL stance, the method that resulted in the largest effect size ($\delta = 0.39$ for DL stance AP spectral content; Table 3). The technique thus appears to show promise for the analysis of COP data in postural stability assessment and may facilitate insight into the effects of pathology on neuromotor control.

Baseline M-BESS scores were comparable with those previously reported in active young adults (Azad et al., 2016; DL/SL/TAN/total mean values = 0.0/1.3/0.3/1.6 compared to 0.0/1.7/0.5/2.2 in current study). We identified an increase in M-BESS scores post-SRC in TAN stance but did not find differences in DL or SL stance. The small post-concussive change observed here is commensurate with previous literature: whilst there have been limited studies

Table 1
Clinician scores for M-BESS. Row highlighted in bold if $p < 0.05$. All values were zero for DL touchdowns so no statistic could be calculated, but can be interpreted as no identified difference.

Stance	Baseline				Post-SRC				W	p	δ
	Mean	Median	Lower quartile	Upper quartile	Mean	Median	Lower quartile	Upper quartile			
DL	0	0	0	0	0	0	0	0	N/A	N/A	–
SL	1.7	1	0.3	2.2	2.1	1.9	0.4	3.4	42.5	0.55	–
TAN	**0.5**	**0**	**0**	**1**	**1.1**	**1**	**0**	**1**	**6**	**0.05**	**0.19**
Total	2.2	1.7	0.5	1	3.3	3	1	4.5	17	0.17	–

Table 2
Discrete point COP variable results: comparison of baseline with post-SRC results in the group obtaining an SRC during the season ($n = 15$). Row highlighted in bold if $p < 0.05$. All values (baseline and post-SRC) were zero for DL touchdowns so no statistic could be calculated, but can be interpreted as no identified difference.

Variable	Stance	SRC baseline				Post-SRC				W	p	δ
		Mean	Median	Lower quartile	Upper quartile	Mean	Median	Lower quartile	Upper quartile			
$Range_{ML}$ (mm)	DL	38.9	38.6	31.3	46.6	44.4	43.5	40.4	46.8	30	0.17	–
	SL	112.2	57.6	52.7	163.7	77.3	56.7	41.2	126.3	83	0.06	–
	TAN	73.8	61.6	50.1	71.9	69.3	61.7	44.0	79.9	48	0.81	–
$Range_{AP}$ (mm)	DL	44.6	41.6	34.2	54.6	49.1	37.6	34.0	52.4	64	0.50	–
	SL	**136.0**	**126.1**	**75.1**	**179.7**	**104.4**	**80.6**	**65.3**	**169.6**	**89**	**0.02**	**0.30**
	TAN	116.8	80.7	57.0	147.1	109.7	100.1	56.0	125.9	60	0.67	–
SD_{ML} (mm)	DL	7.9	8.2	6.4	9.2	9.2	8.2	7.6	9.8	32	0.22	–
	SL	17.7	14.2	11.7	22.4	14.8	12.7	9.3	21.0	79	0.10	–
	TAN	13.7	13.1	11.3	14.3	13.7	13.2	10.2	15.1	51	0.95	–
SD_{AP} (mm)	DL	9.0	8.0	6.7	11.6	9.9	9.0	6.6	12.0	57	0.81	–
	SL	17.5	18.3	14.3	19.0	17.7	14.7	9.9	23.8	81	0.07	–
	TAN	18.8	15.6	11.4	24.0	17.9	15.2	10.1	22.8	61	0.62	–
$COPV_{mean}$ (m s^{-1})	DL	0.02	0.02	0.02	0.02	0.02	0.02	0.02	0.02	48	0.81	–
	SL	0.09	0.06	0.05	0.10	0.07	0.05	0.04	0.07	80	0.09	–
	TAN	0.07	0.05	0.04	0.07	0.06	0.04	0.04	0.06	62	0.58	–
CEA95% (mm^2)	DL	503.8	495.0	329.9	716.2	466.7	357.8	293.1	596.4	78	0.12	–
	SL	3578.5	1478.6	917.3	4231.3	1943.4	702.1	263.0	2416.1	81	0.08	–
	TAN	3135.7	1583.3	798.7	3807.7	2772.9	815.7	435.1	3777.3	57	0.81	–
Touchdowns	DL	0.0	0	0	0	0.0	0	0	0	NA	NA	–
	SL	0.0	0	0	1	0.0	0	0	1	12	1.00	–
	TAN	0.0	0	0	0.8	0.0	0	0	0.8	6	1.00	–

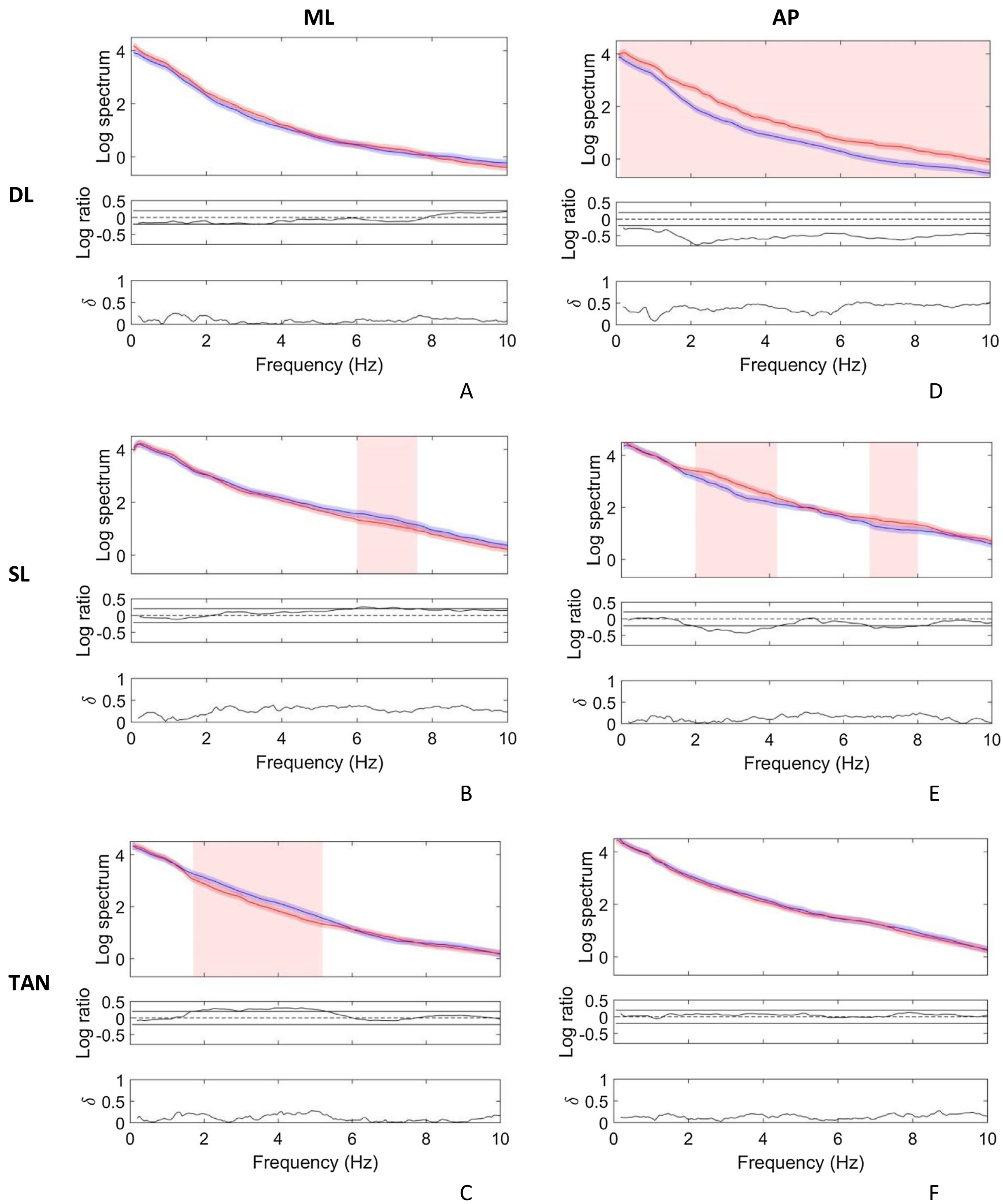

Fig. 2. Spectral analysis plots. (A) DL ML; (B) SL ML; (C) TAN ML; (D) DL AP; (E) SL AP; (F) TAN AP. For each sub-figure, the top panel shows the log pooled spectra for baseline (blue) and post-SRC (red). Dark shaded areas indicate 95% confidence intervals for the pooled spectral estimates. Pale background shading indicates frequency regions where significant differences between baseline and post-SRC spectra were identified. The middle panel shows the null (dotted line) and upper and lower confidence limits (solid horizontal lines) for the log ratio test. The null is rejected for frequency regions in which the log ratio falls outside the confidence limits. The bottom panel shows the pointwise absolute Cliff's delta effect size for the spectral comparison. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

focused on post-concussive M-BESS changes, test scores for the standard BESS (a more-demanding assessment) have been reported to return to baseline levels within 2–5 days after concussion (Murray et al., 2014; Ruhe et al., 2014). No differences in discrete point COP variables were identified between baseline and post-SRC variables in DL and TAN stance. The mean range$_{AP}$ in SL stance decreased post-SRC (δ = 0.30; Table 2). Increases rather than decreases in postural excursion metrics (COP velocity and displacement) are more-commonly reported after concussion, and are traditionally interpreted as indicating impaired postural control (e.g. Baracks et al., 2018; Guskiewicz, 2011; Riemann and Guskiewicz, 2000). There is evidence, however, that postural sway has an

Table 3
Spectral analysis results summary. Row highlighted in bold if p < 0.05. Mean δ is the mean effect size within the band of difference.

Variable		Frequency range (Hz)	Result	Mean δ
DL	ML	–	–	–
	AP	**0.1–10.0**	**Post-SRC > baseline**	**0.39**
SL	**ML**	**6.0–7.6**	**Post-SRC < baseline**	**0.29**
	AP	**2.0–4.2**	**Post-SRC > baseline**	**0.08**
		6.7–8.0	**Post-SRC > baseline**	**0.18**
TAN	**ML**	**1.7–5.2**	**Post-SRC < baseline**	**0.16**
	AP	–	–	–

exploratory role in a healthy neural control system to maintain dynamic input to the central nervous system and to modulate muscle activation (Carpenter et al., 2010; Kiemel et al., 2011; Murnaghan et al., 2013, 2011). Post-SRC decreases in postural excursion metrics have been reported by Hides et al. (2017) and have also been noted in those with neck injuries (Field et al., 2008), which may indicate a reduced ability or willingness to utilise this exploratory function of sway. Further research in the area is required to explain the differing findings to date and elucidate the role of postural sway in healthy and pathological populations.

Changes in spectral content were identified in all three stances after SRC (Table 3). With the exception of DL stance AP content, which increased across the analysed frequency range, changes were localised to the higher frequencies (1.7–8 Hz) considered indicative of proprioceptive system effects (Diener et al., 1984; Fitzpatrick et al., 1992; Mauritz and Dietz, 1980; Nashner, 1976; Paillard and Noé, 2015) and of velocity-based rather than position-based control (Gilfriche et al., 2018). Whilst slowly-adapting peripheral afferent neurons are generally considered sensitive to position, activating relative to the magnitude of a displacement, many rapidly-adapting neurons respond primarily to stimulus velocity rather than displacement (Burgess and Perl, 1973; Esteky and Schwark, 1994; Jeka et al., 2004) and velocity-based control appears to be more accurate for postural control in quiet stance (Jeka et al., 2004; Kiemel et al., 2002). All observed post-SRC increases in spectral content were in the AP direction and all decreases were in the ML direction. An AP increase in 2–20 Hz spectral content has previously been noted following ankle injury (Golomer et al., 1994) and the authors hypothesised that it may relate to a greater contribution from monosynaptic reflexes to maintain balance equilibrium. The AP increase in SL stance was associated with a decrease in ML content, so may represent a redistribution of high-frequency signal energy from the ML to the AP direction.

The greatest effect size and the largest frequency range over which a difference was detected was for the increase in AP spectral content in DL stance. Changes were identified across the investigated frequency range, suggesting potential vestibular and cerebellar integration as well as proprioceptive control system effects of SRC, in concordance with existing literature (Christy et al., 2019; Guskiewicz, 2003; Mallinson and Longridge, 1998; Nowak, 2018; Wright et al., 2017). As all trials were performed with vision obscured, for consistency with M-BESS, identification of any post-SRC effects on visual system integration was outside the scope of the study. DL stance is typically considered to be less sensitive to changes in postural control than SL or TAN in clinical tests where a floor effect is encountered (Hunt et al., 2009), but has been found to be a stronger discriminator of concussed *vs* non-concussed athletes when instrumented outcome measures are analysed, likely because the gross movements required to maintain balance in the more-challenging SL and TAN stances result in greater natural between-trial and between-participant variability (Doherty et al., 2017; King et al., 2017, 2014). Post-SRC changes predominantly in the AP direction have been previously observed in DL stance for other COP variables (Powers et al., 2014) and are likely indicative of alterations in control of the ankle dorsiflexors and plantar flexors, as these are the primary regulators of AP COP movement during quiet standing (Winter, 1995). Future work should focus on further elucidating the mechanisms underpinning these changes.

The most notable limitation of this study is that each stance position was performed only once per session and held for 20 s. Whilst this enabled a direct comparison with the most-common clinician scoring system, it did not allow us to elucidate effects of time on the structure of the COP signal. COP movement is known to exhibit non-stationary characteristics (Carroll and Freedman, 1993) when evaluated over extended periods of time so results should not be extrapolated to the interpretation of longer-duration stances. Effects of repeated testing have been previously reported for BESS, even over time periods of up to 60 days between tests (Valovich McLeod et al., 2004), although no learning effect in the cohort most-closely age matched to ours was observed for either BESS or for COP velocity measures in tests done seven days apart (Alsalaheen et al., 2015). As we did not record whether participants had previous experience of the M-BESS protocol, and performed only a single baseline assessment, we are unable to exclude the possibility that the changes observed were influenced by a learning effect. Further research should therefore focus on investigating longer-term temporal changes in spectral content within a trial and on clarifying the effect of repeated exposure on all outcome measures. Other methods of signal structure analysis, such as entropy, could also be applied alongside this method to compare the results obtained and their sensitivity to pathological changes.

There are known effects of age on postural control COP metrics in both the time domain and the frequency domain (Barozzi et al., 2014; Gouleme et al., 2014; Hugentobler et al., 2016; Quatman-Yates et al., 2018; Singh et al., 2012; Williams et al., 1997) so the spectra and summary statistics presented here should not be generalised to groups of different ages (e.g. young children or older adults). It is unclear whether systematic differences exist within the age range included in our study (Gouleme et al., 2014; Quatman-Yates et al., 2018) and a recent systematic review concluded that athletes from 13 years until the end of adolescence could be treated as a single group as regards concussion treatment and management (Davis et al., 2017).

Our findings indicate that an evaluation of continuous COP frequency spectra can identify post-SRC changes with larger effect sizes than other commonly-utilised metrics, and that the changes identified predominantly occur within sub-bands of the frequency range indicative of peripheral proprioceptive control. Future work should thus focus on higher-resolution elucidation of the mechanisms and pathological indications behind COP trajectory changes within the broad 'high frequency' range in order to facilitate interpretation of the sub-band differences. Reporting and monitoring post-SRC COP changes clinically across the full frequency spectrum may enable specific postural control deficits indicative of neural changes to be identified and monitored after injury and during recovery. A higher-resolution understanding of proprioceptive control mechanisms also has the potential to facilitate individualised clinical rehabilitation programmes based on a patient's identified frequency deficits.

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Declaration of Competing Interest

No conflicts of interest declared.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.109400.

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