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The trunk segmental motion complexity and balance performance in challenging seated perturbation among individuals with spinal cord injury

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Abstract

Background Motion complexity is necessary for adapting to external changes, but little is known about trunk motion complexity during seated perturbation in individuals with spinal cord injury (SCI). We aimed to investigate changes following SCI in trunk segmental motion complexity across different perturbation directions and how they affect postural control ability in individuals with SCI.

Methods A total of 17 individuals with SCI and 18 healthy controls participated in challenging sagittal-seated perturbations with hand protection. Upper arm activation was measured using surface electromyography for trial consistency. Motion complexity parameters, quantified across three degrees of freedom, was assessed using relative angular acceleration from six trunk segments obtained through motion capturing system. Motion performance parameters were assessed using center of pressure (CoP) measurements from a force plate, including settling time, maximum CoP displacement (MD) variability, and steady-state error. Statistical analyses examined group and direction differences, while complexity-performance relationships were evaluated using multiple response least partial squares regression.

Results Compared to healthy controls, individuals with SCI showed significantly lower motion complexity in the lumbar and upper thoracic segments (approximately10% – 20%), with identical settling time and higher MD variability. Backward perturbations, as opposed to forward perturbations, resulted in reduced complexity in the aforementioned segments and increased steady-state error. Lower lumbar rotation complexity negatively correlated with MD variability ($\beta = -0.240$) and steady-state error ($\beta = -0.485$) in individuals with SCI, while showing a minor positive correlation with settling time ($\beta = 0.152$) during backward perturbation.

Conclusion Simplified motion control in individuals with SCI may arise from uncoordinated lumbar and overactive thoracic neuromuscular control, compromising stability despite maintaining speed. Increasing lumbar motion complexity could enhance postural stability and accuracy during backward perturbation, representing a potential target for developing seated balance rehabilitation strategies and promoting more adaptive trunk control.

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Keywords Spinal cord injury, Seated balance, Motion complexity, Perturbation response, Rehabilitation strategy optimization

Background

Spinal cord injury (SCI) is a neurological injury with a high disability rate that necessitates lifelong rehabilitation. Approximately 20.6 million people worldwide are living with SCI, and the incidence has been steadily increasing [1]. About 70–80% of individuals with SCI rely on wheelchairs for mobility, spending around 10 h per day in seated positions [2, 3]. Seated instability often results from trunk neuromuscular weakness under unstable conditions [4]. External perturbations, such as wheelchair strikes, sudden stops, uneven surface and obstacle navigation, significantly increase the risk of falls and can lead to hospitalization [5], highlighting the critical need for effective motion control strategies in rehabilitation.

Normal trunk motion relies on nonlinear interactions among muscle dynamics, joint mechanics, feedback loops, neural processing, and environmental factors, allowing for real-time adaptation to changing surroundings [6]. Therefore, complexity is essential for motion control because nonlinear components introduce unpredictability [7]. Health biological systems, characterized by abundant muscle mass and intricate neural circuitry, maintain extensive coupling relationships that enable flexible use of degrees of freedom (DOF) during tasks, preserving motion complexity [8]. However, pathological conditions or aging can significantly diminish this complexity. Lipsitz and Goldberger proposed the concept of complexity loss, suggesting that diseased states are marked by fewer inputs and connections, resulting simpler output signals [9–11].

Emerging pieces of evidence suggest that both closed-loop and non-closed-loop rehabilitation systems utilizing functional electrical stimulation (FES) can effectively restore trunk stability and enhance daily living performance [12–15]. Close-loop systems typically use accelerometers or electrogoniometers to estimate the forward-leaning trunk angle, allowing for stimulation modulation to trunk extensor muscles based on discrepancies between target and current angles [12, 13]. Non-closed-loop system focus on improving active stiffness through increased co-contractions of trunk muscles in individuals with SCI [14, 15]. Despite enhancing stability and response times, these rehabilitation strategies often prioritize endpoint control over necessary continuous adjustments, sacrificing smooth joint trajectory tracking [16].

Motion complexity shows promise in electrical stimulation rehabilitation systems. However, the theoretical aspect of trunk motion complexity has not yet been explored in seated balance rehabilitation for individuals with SCI. This study aimed to investigate changes in spinal segmental motion complexity and seated perturbation performance following SCI during challenging seated tasks, focusing on their relationship in individuals with SCI. We hypothesized that (1) SCI may reduce spinal segmental motion complexity and impair seated perturbation performance, potentially varying by perturbation direction; and (2) spinal segmental motion complexity offers a comprehensive understanding of motion performance in individuals with SCI, encompassing multiple aspects. The results can potentially inform the development of more effective FES strategies, improving movement responsiveness and enhancing quality of life for individuals with SCI.

Methods

Participants

A power analysis, conducted using G*Power 3.1 software, determined a required sample size of 26 participants. This analysis aimed for 80% power, accounting for 10% dropout rate. The calculation was based on an effect size of 0.614, as reported by Audu et al. (2016) for the relative contribution of derivative component in seated perturbation responses (normal controls: 65±2; individuals with SCI: 57±7) [17].

This study enrolled 17 individuals with SCI and 18 age- and gender-matched healthy controls at Shanghai YangZhi Rehabilitation Hospital (Table 1). Individuals with SCI met the diagnostic criteria for thoracic SCI according to the American Spinal Injury Association for at least one year [18]. They could sit upright for more than 10 min without back support, using their upper extremities for protection. Individuals with SCI were excluded if they had: (1) other neurological illnesses; (2) visual or vestibular impairment; (3) cognitive impairment; and (4) severe complications, such as muscle tone abnormality or neuropathic pain with a VAS score greater than 5. Healthy controls were excluded if they had a history of neurological or trunk musculoskeletal illnesses, or had recently engaged in intense exercise.

This study conformed to the ethical requirements in accordance with the Declaration of Helsinki. Ethical approval for the protocol was obtained from the ethics committee of Shanghai YangZhi Rehabilitation Hospital (SBKT-2024-001). Written informed consent was obtained from each participant prior to the study.

Experimental design and procedure

Participants sat upright in the testing wheelchair on the BalanceTutor (BalanceTutor, MediTouch, USA) after their

Table 1 Participants characteristics

	Control group ($n=18$)	SCI group ($n=17$)	*P* value
Age, years	40.11 ± 12.39	39.59 ± 10.40	0.894
Height, m	1.69 ± 0.06	1.70 ± 0.07	0.771
Weight, kg	67.86 ± 7.70	61.94 ± 10.44	0.064
Gender(male/female), n	11/7	13/4	0.328
Duration of injury, years	NA		NA
> 10		8	
< 10		9	
AISA grade	NA		NA
A		13	
B		2	
C		2	
Injury level	NA		NA
T1-T3		2	
T4-T6		1	
T7-T9		2	
T10-T12		12	

Note Values are mean ± SD or *P* values; ASIA, American spinal injury association

demographic details were recorded (Fig. 1A). They were required to complete challenging seated perturbation tasks, the intensity of which was set to 70% of the maximal acceptable strength in each direction. Participants held on to the wheelchair arm rests for safety, but were instructed to use trunk strength and minimize upper extremity exertion during perturbations. Additionally, a safety monitor will be present during the perturbation process.

Each individual experienced 80 perturbations in total. The first 20 perturbations were conducted at the test intensity before the actual test, serving as the warm-up phase. Sixty perturbations were administered in a sequential condition, divided into four directions (with 15 perturbations each). Forward and backward perturbations were regarded as experimental conditions, whereas leftward and rightward perturbations served as wash-out test conditions. The 60 perturbation tests included three sessions, separated by 3-minute intervals to prevent muscle fatigue. Each session consists of 20 trials, with each trial including one perturbation stimulus and a 15-second rest period. Each perturbation comprised a 300 ms acceleration phase, followed by a brief period of approximately 50 ms during which a relatively constant linear

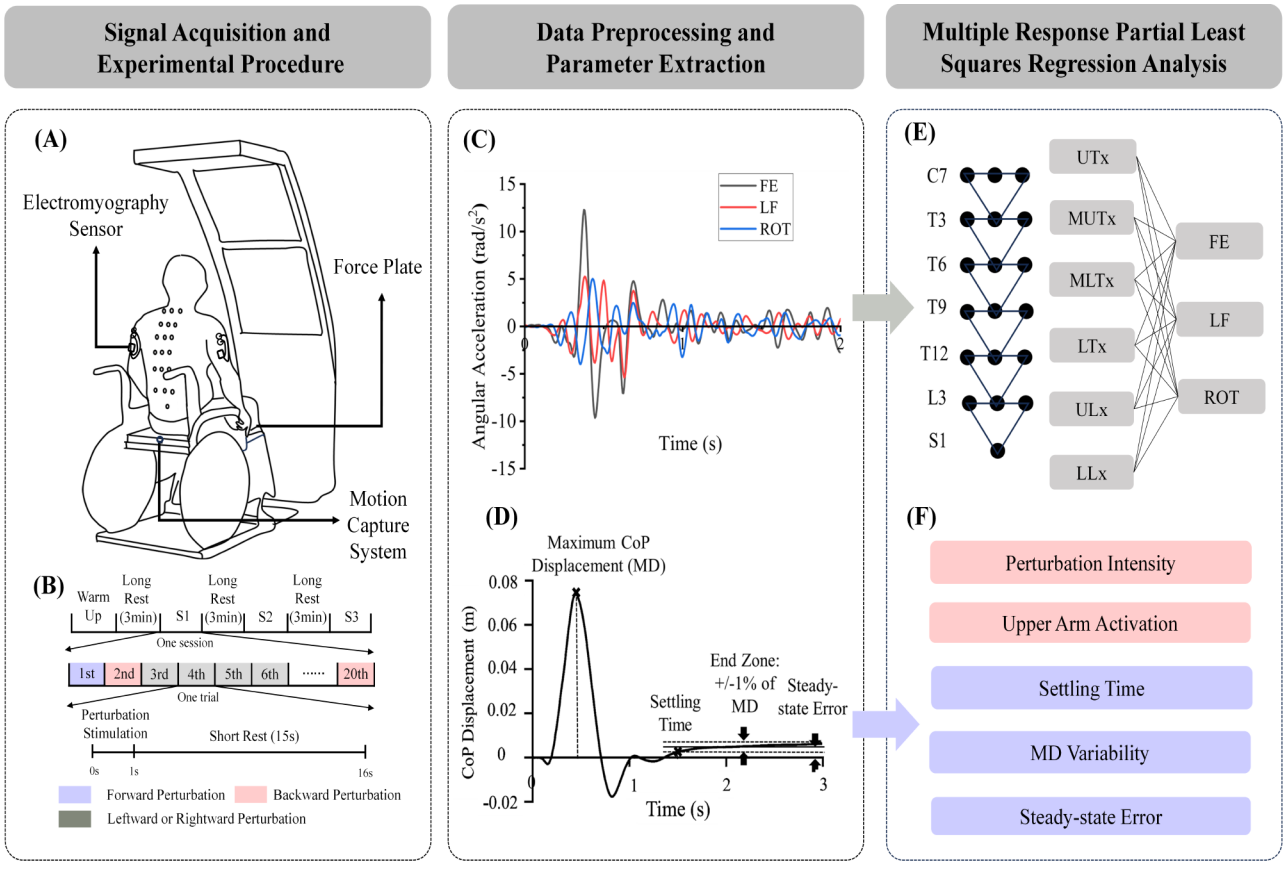

Fig. 1 The experimental design and data processing flow. (**A**) Experimental setup, (**B**) experimental procedure, (**C**) angular acceleration response, (**D**) center of pressure displacement response, (**E**) complexity parameters, and (**F**) seated performance parameters. UTx, upper thoracic segment; MUTx, mid-upper thoracic segment; MLTx, mid-lower thoracic segment; LTx, lower thoracic segment; ULx, upper lumbar segment; LLx, lower lumbar segment; FE, flexion-extension; LF, lateral flexion; ROT, rotation; CoP, center of pressure; MD: maximum center of pressure displacement

velocity was maintained, and concluded with a 300ms deceleration phase.

Data acquisition

Synchronous surface electromyography (sEMG), kinematic, and kinetic data were recorded. Upper arm muscle activation was measured to determine hand force exertion. A wireless sEMG system (Noraxon, Scottsdale, USA) was used to collect data at 2000 Hz from bilateral biceps brachii and triceps brachii. Electrode placement followed the SENIAM recommendation [19]. A Qualisys motion analysis system (Qualisys, Gothenburg, Sweden) captured 3D kinematic data from marker locations at 200 Hz. Marker placement included the spinous processes of the 7th cervical vertebra (C7), the 3rd, 6th, 9th and 12th thoracic vertebrae (T3, T6, T9 and T12), the 3rd lumbar vertebra (L3) and the 1st sacral vertebra (S1). Additionally, markers were also placed 5 cm lateral to C7, T3, T6, T9, T12 and L3. These marker groups defined the following spinal segments: upper lumbar segment (ULx), lower thoracic segment (LTx), mid-lower thoracic segment (MLTx), mid-upper thoracic segment (MUTx) and upper thoracic segment (UTx) [20]. A reflective marker was used to monitor the wheelchair seat motion. Center of pressure (CoP) displacement data was also collected at 200 Hz using a Kistler portable force plate (Kistler, Winterthur, Switzerland) under the buttocks.

sEMG data processing and perturbation quality monitoring

All the data were processed in MATLAB (MathWorks, USA). The raw sEMG data was bandpass filtered (10–400 Hz) with a fourth-order Butterworth filter. The obtained signals were then full-wave rectified, and segmented into 100 ms windows to calculate the root mean square (RMS) of the upper arm. Normalized RMS amplitudes were obtained using the maximum contraction amplitude method [21]. RMS results were averaged, since the study did not focus on the difference between bilateral and muscle-specific activation. Perturbations with outlier RMS values were removed to ensure consistent upper extremity activation levels. These outliers were identified as values exceeding 1.5 times the interquartile range of the RMS values calculated across all perturbations, indicating unusually high levels of upper extremity exertion.

Motion complexity analysis based on kinematic data

A fourth-order Butterworth low-pass filter with an 8 Hz cutoff frequency was applied to trunk segmental motion data [9]. Perturbation initiation was defined as wheelchair acceleration above the baseline mean plus five standard deviations [17]. Relative angular acceleration for each spinal segment was calculated in three DOFs, including flexion-extension (FE), lateral flexion (LF), and rotation (ROT) (Fig. 1C) [20], measuring the orientation of each trunk segment relative to that of the segment below. The Cartesian axis system used to describe different DOFs is consistent with the previous work [20]. Despite the proximity of the spinal segments, the relative angular acceleration shows good resolution and aligns segmental spinal motion patterns with vertebral facet orientations, effectively identifying normal motion trends and detecting differences in individuals with SCI (Fig. S1).

The sample entropy was then used to quantify the complexity of angular acceleration signals during the settling period (the time from the onset of perturbation to reaching settling time), with the Chebyshev distance between each subsequence calculated, followed by counting the number of similar subsequences within the tolerance [22]. The calculation formula was defined as follows:

$$SampEn(m,r) = -In\left(\frac{\sum_{i=1}^{N-1} A_i}{\sum_{i=1}^{N} A_i}\right) \quad (1)$$

Where N represents the length of the time sequence; A_i represents the number of similar subsequences for each subsequence; The subsequence length m was set to 2, and tolerance r was calculated as 0.15 times the standard deviation of the entire time series [23].

Seated perturbation performance analysis based on kinetic data

A fourth-order Butterworth low-pass filter with a 10 Hz cutoff frequency was applied to the CoP data. Three aspects of seated balance ability were assessed based on the CoP data in the perturbation direction (Fig. 1D):

(1) Settling time: It referred to the point at which the fluctuations of CoP first stabilized within 2% of the maximum CoP displacement (MD) for a duration of at least 0.5 s indicating balance recovery speed [24].
(2) MD variability: It was measured by determining the coefficient of variation of max CoP displacement across trials, reflecting postural control stability [25].
(3) Steady-state error: It was the discrepancy between initial and final CoP displacement after response stabilization, quantifying postural control accuracy [24].

Additionally, perturbation intensity and upper arm activation were also considered to be measures of overall motion performance (Fig. 1F).

Statistical analysis

Statistical analysis was performed using SPSS 25 (IBM SPSS Statistics, Chicago, IL, USA). The Shapiro-Wilk test

and Levene's test were used to examine the normality and homogeneity of variance of demographic parameters, seated balance parameters, and trunk motion complexity parameters. Independent samples t-test and chi-square test compared population characteristics and count data between groups. Repeated measures analysis of variance examined the effects of group and perturbation direction on complexity and performance characteristics. Friedman and Dunn-Bonferroni tests assessed motion complexity differences across spinal segments. Pearson correlation coefficients were used to determine the influence of two confounding factors, including intensity and upper arm activation. The statistical significance was set at two-tailed $P<0.05$.

Multiple response partial least squares regression was employed to explore the relationship between trunk motion complexity parameters (18 independent variables, including sample entropy of six spinal segmental motion in three DOFs) and balance performance (5 dependent variables, including speed, stability, and accuracy of seated postural control, perturbation intensity, and upper arm activation) in individuals with SCI, addressing potential multicollinearity. The variables were transformed using Z-score normalization before the analysis. To capture the most relevant information while minimizing overfitting, five components were selected. The model's accuracy was evaluated using leave-one-out cross-validation, as measured by coefficient of determination (R^2) and root mean square error of cross-validation (RMSECV). The regression coefficients indicate the direction and strength of motion complexity's influence on balance performance. The variable importance in projection (VIP) score indicates a significant contribution from specific segmental motion complexity if it is greater than 1.

Results

Trunk segmental motion complexity

Figure 2 depicts the trunk segmental complexity characteristics in all DOFs across different perturbation

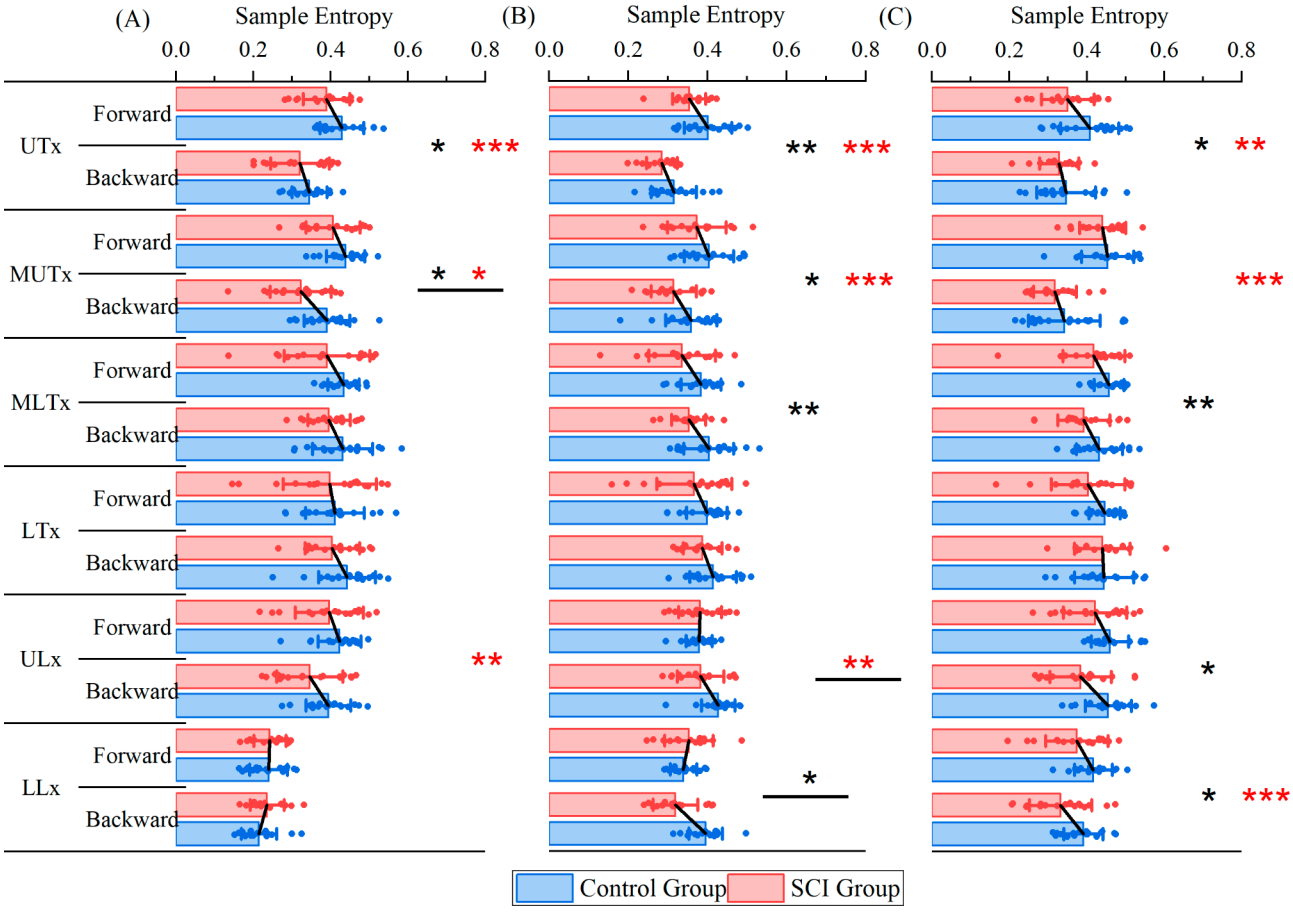

Fig. 2 Sample entropy of trunk segmental motion between groups in (**A**) FE, (**B**) LF, and (**C**) ROT motion. Black asterisks represent significant group main effects, red asterisks represent significant direction main effects, and the black lines under the asterisks represent interaction effects. The black lines connecting two bars show the mean complexity levels between groups. *: $P<0.05$, **: $P<0.01$, ***: $P<0.001$. UTx, upper thoracic segment; MUTx, mid-upper thoracic segment; MLTx, mid-lower thoracic segment; LTx, lower thoracic segment; ULx, upper lumbar segment; LLx, lower lumbar segment; FE, flexion-extension; LF, lateral flexion; ROT, rotation

Table 2 Motion complexity within different segments

	Direction	DOF	Spine Segment						*P* value
			UTx	MUTx	MLTx	LTx	ULx	LLx	
Control group	Forward	FE	0.429 ± 0.055 [e]	0.439 ± 0.048 [i]	0.433 ± 0.039 [l]	0.411 ± 0.074 [n]	0.423 ± 0.054 [o]	0.239 ± 0.047	< 0.001***
		LF	0.401 ± 0.058 [e]	0.403 ± 0.060 [i]	0.384 ± 0.049	0.398 ± 0.049 [n]	0.378 ± 0.032	0.339 ± 0.033	< 0.001***
		ROT	0.407 ± 0.073	0.454 ± 0.065	0.457 ± 0.038 [l]	0.446 ± 0.040	0.460 ± 0.046	0.417 ± 0.047	0.021*
	Backward	FE	0.345 ± 0.044 [b, c]	0.391 ± 0.057 [i]	0.431 ± 0.075 [l]	0.442 ± 0.071 [n]	0.394 ± 0.056 [o]	0.215 ± 0.044	< 0.001***
		LF	0.315 ± 0.056 [b, c, d, e]	0.359 ± 0.063 [h]	0.403 ± 0.061	0.414 ± 0.057	0.427 ± 0.041	0.395 ± 0.041	< 0.001***
		ROT	0.346 ± 0.074 [b, c, d]	0.342 ± 0.090 [f, g, h]	0.432 ± 0.057	0.444 ± 0.074	0.455 ± 0.058 [o]	0.391 ± 0.049	< 0.001***
SCI group	Forward	FE	0.389 ± 0.059 [e]	0.406 ± 0.068 [i]	0.391 ± 0.107 [l]	0.397 ± 0.117 [n]	0.396 ± 0.085 [o]	0.243 ± 0.040	< 0.001***
		LF	0.354 ± 0.040	0.373 ± 0.072	0.336 ± 0.082	0.366 ± 0.092	0.381 ± 0.053	0.353 ± 0.059	0.162
		ROT	0.350 ± 0.066 [a, d]	0.440 ± 0.058	0.418 ± 0.077	0.403 ± 0.092	0.421 ± 0.079	0.374 ± 0.079	0.001**
	Backward	FE	0.321 ± 0.073 [e]	0.323 ± 0.076 [i]	0.395 ± 0.053 [l]	0.404 ± 0.068 [n]	0.346 ± 0.083 [o]	0.235 ± 0.043	< 0.001***
		LF	0.285 ± 0.037 [b, c, d]	0.314 ± 0.055 [g, h]	0.353 ± 0.042	0.387 ± 0.047 [n]	0.383 ± 0.057 [o]	0.319 ± 0.055	< 0.001***
		ROT	0.328 ± 0.049 [c]	0.318 ± 0.054 [g]	0.392 ± 0.065	0.440 ± 0.070 [n]	0.384 ± 0.077	0.332 ± 0.078	< 0.001***

Note Values are mean ± SD or *P* values

a, b, c, d, e indicates that UTx is statistically different from MUTx, MLTx, LTx, ULx and LLx, respectively. f, g, h, i indicates that MUTx is statistically different from MLTx, LTx, ULx, and LLx, respectively. j, k, l indicates that MLTx is statistically different from LTx, ULx, and LLx, respectively. m, n indicates that LTx is statistically different from ULx and LLx, respectively. o indicates that ULx is statistically different from LLx. * $P<0.05$; ** $P<0.01$; *** $P<0.001$. UTx, upper thoracic segment; MUTx, mid-upper thoracic segment; MLTx, mid-lower thoracic segment; LTx, lower thoracic segment; ULx, upper lumbar segment; LLx, lower lumbar segment; DOF, degree of freedom; FE, flextion-extension; LF, lateral flexion; ROT, rotation

Table 3 Performance parameters in seated perturbation

	Direction	Subjects		Statistics (*P* value)		
		Control group	SCI group	Subjects	Directions	Interaction
Intensity (m/s^2)	Forward	2.922 ± 0.143	2.582 ± 0.197	< 0.001***	0.001**	0.580
	Backward	2.723 ± 0.393	2.434 ± 0.266			
Upper arm activation (% max)	Forward	8.477 ± 3.259	11.586 ± 4.395	< 0.001***	0.141	0.004**
	Backward	7.227 ± 2.526	15.813 ± 5.595			
Settling time (s)	Forward	1.789 ± 0.113	1.701 ± 0.354	0.752	0.342	0.175
	Backward	1.786 ± 0.212	1.842 ± 0.252			
MD variability (%)	Forward	8.138 ± 3.093	11.062 ± 5.116	0.005**	0.097	0.411
	Backward	9.499 ± 3.038	14.442 ± 9.174			
Steady-state error (m)	Forward	0.011 ± 0.033	0.007 ± 0.024	0.003**	< 0.001***	0.307
	Backward	0.015 ± 0.056	0.010 ± 0.061			

Note Values are mean ± SD or *P* values; *: $P<0.05$; **: $P<0.01$; ***: $P<0.001$. MD, maximum center of pressure displacement

directions and groups. Across different groups, SCI group showed lower trunk motion complexity than healthy controls, mainly in the upper thoracic and lumbar segments, including UTx in three DOFs (FE: F = 5.484, $P=0.025$; LF: F = 10.008, $P=0.003$; ROT: F = 4.439, $P=0.043$), MUTx in two DOFs (FE: F = 4.176, $P=0.049$; LF: F = 4.473, $P=0.042$), MLTx in two DOFs (LF: F = 9.238, $P=0.005$; ROT: F = 7.959, $P=0.008$), ULx ROT (F = 7.572, $P=0.010$), and LLx in two DOFs (LF: F = 5.126, $P=0.030$; ROT: F = 7.239, $P=0.011$). Notably, forward perturbations led to a 10.212% decrease in LLx ROT motion and an 8.642% decrease in ULx ROT motion in SCI group, while backward perturbations resulted in a 19.389% decrease in LLx LF motion, a 15.025% decrease in LLx ROT motion, and a 15.657% decrease in ULx ROT motion. This suggests simplified trunk movement patterns and increased trunk stiffness in individuals with SCI.

Regarding different directions, the trunk motion complexity was lower in backward perturbation than forward perturbation. The distinct spinal segments were consistent across perturbation directions and groups, including UTx in three DOFs (FE: F = 24.491, $P<0.001$; LF: F = 42.490, $P<0.001$; ROT: F = 7.866, $P=0.008$), MUTx in three DOFs (FE: F = 7.527, $P=0.01$; LF: F = 16.186, $P<0.001$; ROT: F = 57.786, $P<0.001$), ULx in two DOFs (FE: F = 9.117, $P=0.005$; LF: F = 8.231, $P=0.007$), and LLx ROT (F = 23.922, $P<0.001$).

Table 2 provides a detailed comparison of complexity data across various trunk segments. Notably, the motion complexity in the LLx and upper thoracic segments was simultaneously lower across all three DOFs during the backward perturbation, which indicate that motion flexibility of trunk segments was more significantly constrained in backward perturbations.

Seated perturbation performance

Table 3 summarizes the results of seated perturbation performance across groups and directions. Across

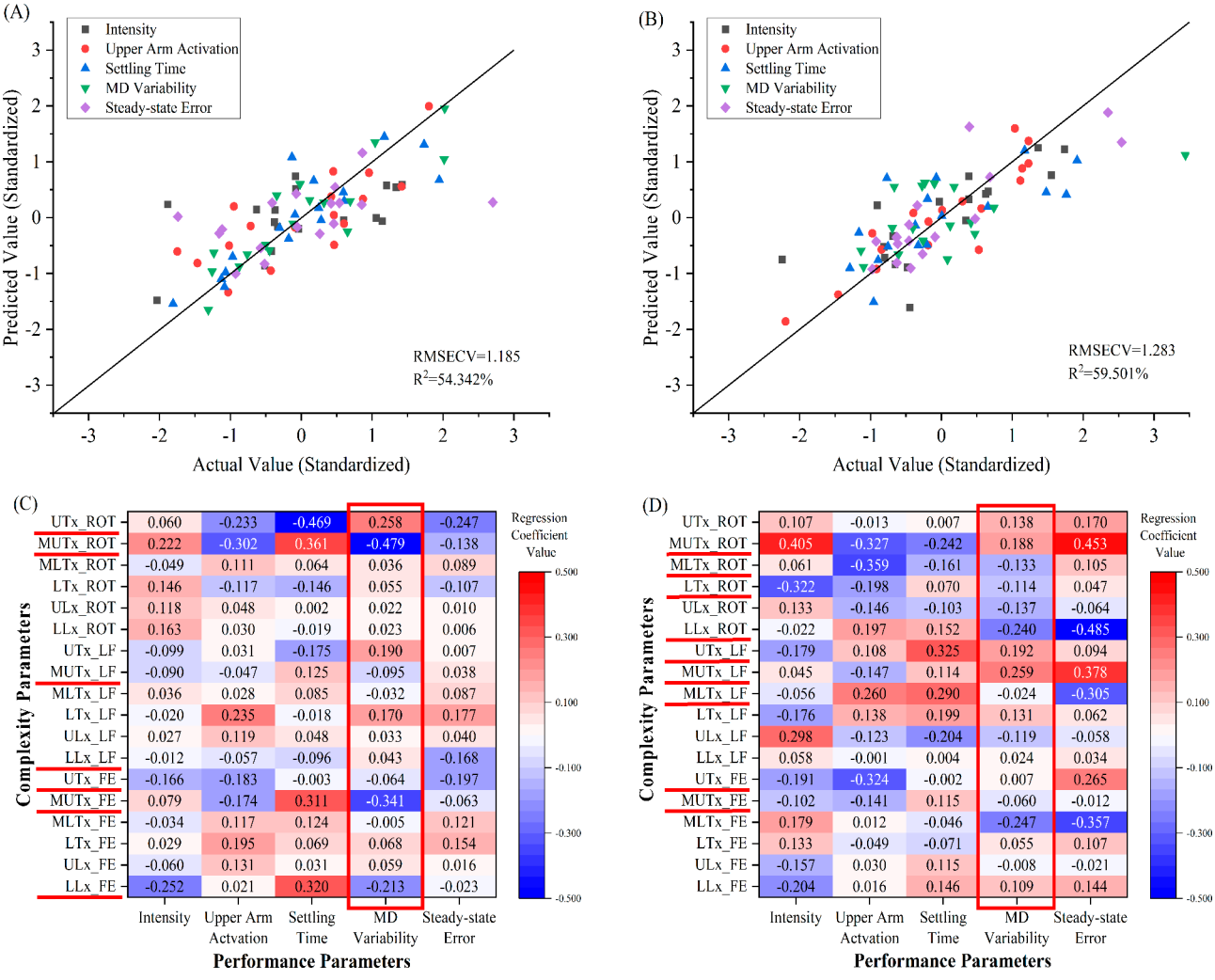

Fig. 3 Multiple response partial least squares regression analysis results in SCI patients, including model performance in (**A**) forward and (**B**) backward perturbations, and regression coefficient matrix in (**C**) forward and (**D**) backward perturbations. The indices indicated by the red lines in (**C**) and (**D**) represent the complexity indices with VIP values greater than 1. UTx, upper thoracic segment; MUTx, mid-upper thoracic segment; MLTx, mid-lower thoracic segment; LTx, lower thoracic segment; ULx, upper lumbar segment; LLx, lower lumbar segment; FE, flextion-extension; LF, lateral flexion; ROT, rotation; MD, maximum center of pressure displacement; VIP, variable importance in projection; RMSECV, root mean square error of cross-validation; R^2, coefficient of determination

different groups, SCI group exhibited reduced perturbation intensities (F=16.188, $P<0.001$), higher upper arm activation (F=32.655, $P<0.001$), and higher MD variability (F=9.047, $P=0.005$) than normal people, indicating poor balance control. However, no significant group differences were observed in settling time, and steady-state error was significantly lower in SCI group (F=10.093, $P=0.003$), showing the favorable aspect of balance control.

Regarding different directions, participants experienced significantly lower intensity (F=14.620, $P=0.001$) and higher steady-state error (F=16.394, $P<0.001$) in backward perturbation, indicating poorer performance in backward perturbation. There was a significant interaction effect between different groups and directions in upper arm activation (F=9.519, $P=0.004$).

Relationship between motion complexity and performance

Figure 3 depicts the outcome of multiple response partial least squares regression analysis of spinal segmental complexity and perturbation performance in SCI group. Complexity variations account for 54.342% and 59.501% of the variance in the forward and backward perturbation performance, respectively, which suggests a moderate level of explanatory power of complexity on performance (Fig. 3A and B).

In the forward perturbation, the indices having the most negative influence on predicting MD variability with VIP above 1 primarily include MUTx ROT (β =

-0.479, VIP=1.623), MUTx FE (β = -0.341, VIP=1.224), and LLx FE (β = -0.213, VIP=1.066). These indices were also positively correlated with settling time (MUTx ROT: β=0.361; MUTx FE: β=0.311; LLx FE: β=0.320). This suggests decreased complexity in these segments may cause postural instability, while maintaining balance recovery speed (Fig. 3C).

During backward perturbation, LLx ROT (β = -0.240, VIP=1.075), MLTx ROT (β = -0.133, VIP=1.117), and LTx ROT (β = -0.114, VIP=1.078) had the most negative influence on predicting MD variability with VIP greater than 1. LLx ROT showed a positive correlation with settling time (β=0.152) and a negative correlation with steady-state error (β = -0.485). In addition to the trade-off between control speed and stability, this also suggests that decreased complexity may negatively affect accuracy (Fig. 3D).

Discussion

This study comprehensively examined trunk segmental motion complexity in individuals with SCI during challenging seated perturbations, quantifying changes across six trunk segments (UTx, MUTx, MLTx, LTx, ULx, and LLx) and analyzing their influence on seated postural control parameters, specifically speed, stability, and accuracy. A thorough understanding of these changes is essential for optimizing rehabilitation strategies, particularly for FES system development, due to the inherent role of motion complexity in adaptive responses to unpredictable environmental events.

We discovered that motion complexity predominantly exhibited a decreasing trend in individuals with SCI compared to healthy controls. This is consistent with previous research on complexity changes during challenging postural tasks due to ageing [26] and other medical conditions [27], as well as simulated outcomes on seated perturbations in individuals with SCI [17], suggesting a simplified motion control. The reduced complexity was identified in the upper thoracic and lumbar regions, stemming from different underlying mechanisms. In the lumbar regions, the decreased complexity in LF and ROT, with a slight, non-significant increase in FE, was likely driven by impaired and uncoordinated neuromuscular control. This is attributed to reduced cortical inhibition of antagonistic muscles and increased muscle co-activation after SCI, particularly between RA and ES [28, 29]. Therefore, lumbar motion flexibility in the non-perturbation direction may be compromised. For the upper thoracic region, especially UTx, the reduced complexity across all DOFs may be due to excessive muscle activation involving non-paralyzed postural muscles like the latissimus dorsi and trapezius [30]. This increased stiffness seems to compensate for inadequate neuromuscular control in the lumbar region.

Additionally, we observed that backward perturbations presented greater challenges. The upper thoracic and lumbar segments, known for their range of motion in FE and ROT motion, respectively, displayed significantly reduced motion complexity across all three DOFs during backward perturbations. This simultaneous locking across all DOFs was not observed during forward perturbations [31]. Furthermore, individuals with SCI exhibit reduced motion complexity in the ULx and LLx segments in LF motion, suggesting increased difficulty. The strong proprioceptive demands, evidenced by higher steady-state error in all participants during backward perturbation, might contribute to this challenge. This is possibly due to the forward head tilt induced by backward perturbation, which may cause greater visual interference. Another possible explanation is the intricacy of extensor neuromuscular control, along with a greater impairment of extensor function in individuals with SCI. In fact, paraspinal muscles are predominantly innervated unilaterally, while the abdominal muscles receive bilateral innervation [32]. Moreover, the different muscle fascicles of the paraspinal muscles are represented at distinct locations within the motor cortex [33]. Additionally, most participants in this study have complete SCI at the T9-T12 segments, resulting in greater damage to their lumbar ES than to the RA. Considering the fact that accidents related to wheelchairs often mainly involve forward falls [34], backward perturbations hold significant clinical relevance.

Individuals displayed a wide range of performance characteristics across several domains of postural control as a result of the rigid motion strategy. In individuals with SCI, we found similar settling time and lower steady-state error compared with healthy controls. This is similar to previous studies, which demonstrated that increased trunk or leg stiffness can enhance CoP movement speed without affecting CoP fluctuation [14, 35]. However, this strategy might not be the most effective control strategy [36]. Individuals with SCI still experienced motion performance limitations, which manifested as lower perturbation intensity, greater reliance on hand support, and greater MD variability. This finding is consistent with the results in a similar unstable seated balancing task [16]. The instability could be produced by an increase in signal-dependent noise as trunk stiffness increases, implying that the system could be prone to internal perturbation caused by the recruitment of more motor units. This would increase muscle force variability while reducing the system's stability and coordination [16].

When investigating the complexity-performance relationship, the capacity to anticipate MD variability is the most critical element, primarily expressed in the upper thoracic and lumbar segments. Reduced complexity in the upper thoracic region can have various implications

in different directions. In backward perturbations, individuals with SCI rely more on hand-assisted strategies, consistent with their higher hand assistive forces. However, stiffer upper thoracic segments may limit the ability to generate more trunk flexion torque in forward perturbation. As a compensatory strategy, the changes in the upper thoracic segments are mostly caused by impaired lumbar neuromuscular control. We further found that the complexity of LLx FE and LLx ROT was adversely connected with stability during forward and backward perturbations, respectively. Reduced LLx complexity may produce improved postural control speed at the expense of stability, but it may also compromise accuracy for backward perturbations. The decreased lumbar rotation complexity is associated with decreased pelvic motion, which may impede the ability to alter the center of mass and jeopardize postural stability [37, 38]. Furthermore, this reduction may also degrade the ability to respond to multi-directional body shift arising from impaired proprioception following SCI [39].

Therefore, appropriately increasing motion complexity, particularly in LLx, is an important factor contributing to adaptability to physical stressors in challenging tasks [40]. This means that individuals with SCI can activate paralyzed muscles through FES, improving neural control and expanding the number of available movement strategies. Additionally, motion complexity comprehensively considers various performance aspects, especially response speed and accuracy, making it a suitable target for optimizing FES strategies. In a closed-loop system, a dual feedback approach can be implemented, whereby motion complexity is optimized as a feedback target in addition to tracking movement angles. This facilitates precise and adaptive control of muscle stimulation. In non-closed-loop systems, motion complexity can serve as a focal point for further optimizing the coordination between trunk agonist and antagonist muscles, leading to smoother movements.

Limitations should be acknowledged. Firstly, the experimental setup had participants consistently holding onto the wheelchair handles while the perturbation intensity was set at a relatively maximum tolerable level. This approach aimed to ensure the safety of individuals with SCI while observing their control strategies under extreme seated balance perturbations. As there was little observed statistical correlation between perturbation intensity (or upper arm activation) and other parameters in this study, the adoption of this experimental paradigm is reasonable to some extent (Table S1). Secondly, this study focused on high-functioning individuals with SCI due to the difficulty low-functioning individuals experience with perturbation tasks, even with hand assistance. In this study, the motion complexity characteristics of healthy individuals also provide a valuable reference for low-functioning individuals, with the aim of reducing fall risks. Future research should investigate the specific characteristics of trunk motion complexity in low-functioning individuals, as well as consider the impact of injury completeness and other factors influencing functional residuals.

Conclusion

In challenging seated perturbation, the simplified motion control observed in individuals with SCI may indicate uncoordinated lumbar and excessive thoracic neuromuscular control, which can assist individuals with SCI in rapid postural control but cannot compensate for the deficiencies in stability. Appropriately increased motion complexity, such as in lumbosacral rotation, can improve stability, especially for backward perturbation. Motion complexity holds promise as a target for optimizing seated balance rehabilitation strategies, promoting more natural and effective trunk control.

Abbreviations

CoP	Center of pressure
DOF	Degree of freedom
FE	Flexion-extension
FES	Functional electrical stimulation
LF	Lateral flexion
LLx	Lower lumbar segment
LTx	Lower thoracic segment
MD	Maximum center of pressure displacement
MLTx	Mid-lower thoracic segment
MUTx	Mid-upper thoracic segment
R^2	Coefficient of determination
RMS	Root mean square
RMSECV	Root mean square error of cross-validation
ROT	Rotation
SCI	Spinal cord injury
sEMG	Surface electromyography
ULx	Upper lumbar segment
UTx	Upper thoracic segment
VIP	Variable importance in projection

Supplementary Information

The online version contains supplementary material available at https://doi.org/10.1186/s12984-024-01522-7.

Supplementary Material 1: Figure S1. Average relative angular acceleration values for each spinal segment during (a) FE, (b) LF and (c) ROT motion in forward perturbation and (d) FE, (e) LF, (f) ROT motion in backward perturbation. Table. S1 Correlations between perturbation intensity/ upper arm activation, and other parameters.

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Author contributions

M.S.: Conceptualization, Methodology, Software, Formal analysis, Investigation, Data curation, Visualization, Writing - original draft; C.L.: Investigation, Data curation; J.S.: Investigation, Data curation; H.X.: Conceptualization, Patients organization; Y.Q.: Patients organization, Project administration, Writing - review and editing, Funding acquisition; W.N.: Conceptualization, Writing

- review and editing, Project administration, Funding acquisition, Supervision. M.Z.: Writing - review and editing, Supervision.

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Data availability
No datasets were generated or analysed during the current study.

Declarations

Human ethics and consent to participate
The study was conducted in accordance with the Declaration of Helsinki and was approved by the ethics committee of Shanghai YangZhi Rehabilitation Hospital (SBKT-2024-001). All participants provided written informed consent prior to participation.

Consent for publication
Consent for publication were given by all participants.

Competing interests
The authors declare no competing interests.

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