2021 43rd Annual International Conference of the
IEEE Engineering in Medicine & Biology Society (EMBC)
Oct 31 - Nov 4, 2021. Virtual Conference

Balance Control Strategies during Perturbed Standing after a Traumatic Brain Injury: Kinematic Analysis*

Naphtaly Ehrenberg, Akhila Veerubhotla, *Member, IEEE,* Karen Nolan, *Member, IEEE*, and Rakesh Pilkar, *Member, IEEE*

***Abstract*— The objective of the current investigation was to examine the presence, absence or alteration of fundamental postural control strategies in individuals post traumatic brain injury (TBI) in response to base of support perturbations in the anterior-posterior (AP) direction. Four age-matched healthy controls (age: 46.50 ± 5.45 years) and four individuals diagnosed with TBI (age: 48.50 ± 9.47 years, time since injury: 6.02 ± 4.47 years) performed standing on instrumented balance platform with integrated force plates while 3D motion capture data was collected at 60 Hz. The platform was programmed to move in the AP direction, during a sequence of 5 perturbations delivered in a sinusoidal pattern at a frequency of 1 Hz, with decreasing amplitudes of 10, 8, 6, 4, and 2 mm respectively. The sagittal plane peak-to-peak range and root mean square (RMS) of the hip, knee, and ankle joint angles during the 5 seconds of perturbation were computed from optical motion capture data. The TBI group had a higher mean range (5.17 ± 1.91°) about the ankle compared to the HC group (4.17 ± 0.81°) for the 10mm perturbation, but their mean range was smaller than the HCs for the other 4 conditions. About the hip, the TBI group's mean range was larger than the HC's for all conditions. For both groups, the mean range decreased with perturbation amplitude for all conditions. The TBI group showed larger changes in mean range and RMS values as the amplitude of the perturbation changed, while the HC group showed smaller inter-trial changes. The results suggest that the TBI group was substantially more reliant on the hip strategy to maintain balance during the perturbations and this reliance was well linked with perturbation amplitude.**

***Clinical Relevance*— Existing information regarding changes in postural control strategies in individuals post TBI is limited. The current work demonstrates lower limb kinematic differences between HC and TBI and some preliminary evidence on increased hip movement in the TBI group.**

I. Introduction

Postural control, the neuromechanical process of maintaining the body's center of gravity (COG) over its base of support, is accomplished through the integration and coordination of visual, vestibular, and somatosensory inputs, and motor control [1]. Postural control is an essential ability that plays a crucial role in many everyday activities, as good balance is required for the efficient execution of functional movements during activities of daily living [2]. A traumatic brain injury (TBI) can cause damage to diffuse areas of the brain, resulting in some or all of the neuromechanical components of balance being affected by the injury. TBI affects the ability to accurately perceive and integrate the relevant sensory information required to be aware of the status of the body [3, 4]. It also affects the ability to properly coordinate or generate the neuromuscular outputs needed to maintain balance during static and dynamic conditions, thus substantially affecting functional activites [2]. To determine the optimal rehabilitation treatment strategies post TBI, it is important to be able to properly assess the balance function of individuals with a given type of injury. However, this is particularly challenging in individuals post TBI due to the wide array and complexity of symptoms that can occur as a result of their injury [2]. While there are many well-established methods for testing postural function that have been validated in healthy individuals and individuals with a variety of diagnoses, they predominantly focus on center of pressure (COP) metrics only [5]. The evidence on kinematic mechanisms in association with balance impairment in individuals post TBI is limited [6]. The lower limb kinematics of postural control are typically referred to as postural strategies, and are categorized by the movement about the joint or joints involved in maintaining balance within each strategy [1, 7, 8]. Two popular strategies that involve only a single joint are the ankle strategy and the hip strategy, while a third combination strategy that involves movement about two or more lower limb joints is also often used [7, 9, 10]. Postural control strategies are critical for maintaining balance as they represent the body's anticipatory and compensatory responses to self-initiated within-body perturbations (standing, reaching, weight shifts) or external perturbations (slips, base of support changes, etc.). The capabilities of each postural control strategy are governed by the mechanical constraints of the joints involved and the body's ability to correctly detect COG errors [1]. As a result, sensory deficits can affect an individual's ability to use the simplest and most efficient strategy for a given situation due to their inability to properly detect COG errors [1], and by extension, motor control deficits such as muscle weakness can also affect an individual's ability to execute a preferred strategy. Research shows that healthy individuals primarily employ the ankle strategy during small

*Research supported by the New Jersey Commission on Brain Injury Research (NJCBIR) Multi-Investigator Grant CBIR15MIG004.

Naphtaly Ehrenberg is with the Center for Mobility and Rehabilitation Engineering Research (CMRER) at Kessler Foundation, West Orange, NJ 07052 USA (e-mail: nehrenberg@kesslerfoundation.org).

Akhila Veerubhotla is with the Dept. of Physical Medicine & Rehabilitation, Rutgers- New Jersey Medical School (NJMS) and Kessler Foundation, West Orange, NJ 07052 USA (e-mail: aveerubhotla@kesslerfoundation.org).

Karen Nolan is with the CMRE, Kessler Foundation, West Orange, NJ 07052 USA, Rutgers NJMS, Newark, NJ, and Children's Specialized Hospital, New Brunswick, NJ (e-mail: knolan@kesslerfoundation.org).

Rakesh Pilkar is with the CMRER at Kessler Foundation, West Orange, NJ 07052 USA and Rutgers NJMS, Newark, NJ (phone: 973-243-6838; fax: 973-324-3527; e-mail: rpilkar@kesslerfoundation.org).

perturbations, and the hip strategy during more significant perturbations [10]. Whether these preferred strategies are altered after a TBI or are replaced by other combinatory strategies prefered by individuals post TBI is unclear [6].

The objective of the current investigation was to examine the presence, absence or alteration of the fundamental postural control strategies in individuals post TBI in response to base of support perturbations in the anterior-posterior (AP) direction. We hypothesized that the individuals post TBI will show either a larger range of movement about their ankles, or motion about both their hips and ankles in a combined strategy, to achieve sufficient postural control during perturbations, compared to healthy controls (HC). The rationale for our hypothesis is that individuals post TBI may have insufficient muscle strength to deploy an effective ankle strategy. Further, sensory perception deficits combined with motor impairment post TBI may result in a delayed perception of perturbations and insufficient execution of the ankle strategy, as the COM adjustment needed to maintain stability may exceed the limits of the ankle joint.

II. Methods

A. Participants

The Institutional Review Board approved all study procedures escribed in this paper. Individuals with chronic TBI (n=4) and aged matched healthy controls (HC) (n=4) were recruited for participation. All individuals: 1) were between 18 and 65 years of age; 2) were not planning any drastic medication changes for at least 2 months; and 3) were able to stand unsupported for at least five minutes; 4) had no history of injury to the lower limbs within the past 90 days; 5) had no severe cardiac disease such as heart attack or moderate or severe congestive heart failure; 6) had no orthopedic, neuromuscular, or neurological conditions that would interfere with their movement; 7) did not currently take any medications that affect balance, strength, or muscle coordination. Additionally, individuals with TBI: 1) were diagnosed with a non-penetrating TBI, six or more months prior to study participation; 2) were medically stable for at least three months prior to their most recent TBI; 3) did not have any uncorrected visual impairment that affected their ability to stand or see a distance of five to ten feet; 4) had not previously been diagnosed with balance dysfunction prior to their TBI. All procedures were approved by the institutional review board, and informed consent was obtained from all participants prior to participation.

B. Study procedures

After obtaining consent, participants performed clinical assessments of balance and mobility. Assessments included the berg balance scale (BBS), timed up and go (TUG), 5-meter walk test (5MWT), and 10-meter walk test (10MWT). After completing the clinical assessments, the participants put on a safety harness and study specific sneakers (individually sized and fit) (New Balance M/WW575VW). Then, 14-mm spherical retroreflective markers were affixed to specific joints and anatomical landmarks according to a modified Helen-Hayes marker-set [11] for full-body motion capture. Participants were asked to stand with their eyes open and their feet properly positioned on the NeuroCom Smart Equitest Clinical Research System (CRS) (Natus Medical Inc. Pleasanton, CA). The NeuroCom is a balance platform with integrated force plates that can be programmed to move in the AP direction (Fig. 1). The safety harness attached to NeuroCom's overhead support beam was adjusted to provide sufficient slack to allow the participants' sufficient range of motion for the upcoming tasks, while also ensuring their safety (Fig. 1). The NeuroCom platform was then used to provide precise preprogrammed perturbations to the participant's base of support in a sinusoidal motion at a rate of 1 Hz over the course of 5 trials of 15 seconds each (Fig. 1). The peak-to-peak perturbation amplitudes used were 10, 8, 6, 4, and 2 mm (high to low order), respectively. Each trial consisted of five seconds of quiet standing, five seconds of perturbation, followed by an additional five seconds of quiet standing (Fig. 2).

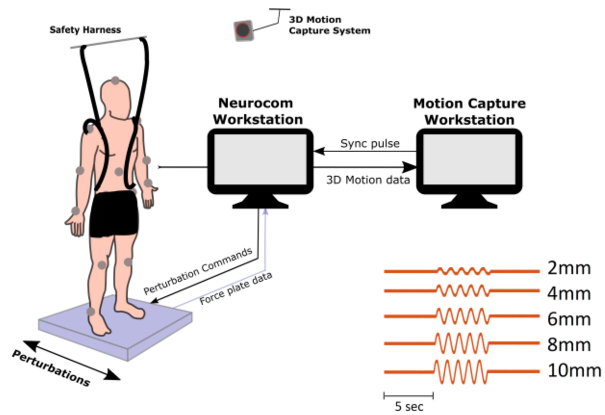

Figure 1. Experimental set up with the platform perturbation profiles.

Optical motion capture data was collected at 60 Hz using Motion Analysis Corporation's (MAC, Santa Rosa, CA) Cortex software and 10 Kestrel 2200 motion capture cameras. Kinematic data was exported from Cortex and processed using a custom MATLAB (Mathworks Inc, Natick, MA) script to calculate the peak-to-peak range and root mean square (RMS) of the hip, knee, and ankle joint angles in the sagittal plane during the 5 seconds of perturbation during each trial. Statistical analysis were performed in SPSS version 26 (IBM Corp., Armonk, NY, USA)

C. Statistical analysis

Descriptive statistics and box plots were used to identify any outliers in the data. The Shapiro-Wilk test was used to test the normality of the dependent variable (mean range of joint angles) for all perturbations and joints for both the HC and TBI groups. The instances where normality was not met ($p<0.05$) due to the small sample size of the study, the skewness and kurtosis was used to determine if the outcome had near-normal distribution. Maulchy's test of Sphericity was used to test the sphericity for both of the within-subject factors (joints and platform perturbations). A mixed methods ANOVA was conducted with group (TBI, HC) as the between subjects factor and joints (hip, knee and ankle), and perturbation amplitudes (10mm, 8mm, 6mm, 4mm, 2mm) as the within-subject factors. The significance level was set to 95% for all tests.

III. Results

Demographics and functional assessments of the HC and the TBI groups are presented in Table 1. Lower performance on the functional assessments (BBS, TUG, 5MWT and 10MWT) demonstrated impaired static and dynamic balance and mobility for the TBI group.

TABLE I. PARTICIPANT DEMOGRAPHICS

ID	Severity	Age	Height (cm)	Weight (kg)	TSI (yrs)	BBS	TUG (sec)	5MWT (sec)	10MWT (sec)
HC1	-	53	177.17	68.95	-	56	2.77	5.62	7.28
HC2	-	40	173.99	67.36	-	56	2.39	4.74	7.87
HC3	-	45	170.18	116.57	-	56	3.27	6.08	7.63
HC4	-	48	154.94	86.18	-	56	3.64	6.61	6.88
	Mean	**46.50**	**169.07**	**84.77**		**56**	**3.02**	**5.76**	**7.42**
	SD	5.45	9.84	22.85		0.00	0.55	0.79	0.43
TBI1	Mild	49	172.72	66.68	1.67	46	4.77	9.63	16.10
TBI2	Moderate*	54	156.21	73.48	2.70	52	4.32	8.16	10.12
TBI3	Mild	56	166.37	119.29	10.34	55	4.25	8.24	11.84
TBI4	Moderate*	35	182.88	69.40	9.38	44	3.50	6.02	11.54
	Mean	**48.50**	**169.55**	**82.21**	**6.02**	**49.25**	**4.21**	**8.01**	**12.40**
	SD	9.47	11.19	24.88	4.47	5.12	0.52	1.49	2.58

* Moderate/Severe

A. *Lower Limb Kinematics*

Fig. 2 shows the mean range and RMS of movement about the lower limb joints for both the HC and TBI groups across all conditions. The HC group's mean range about the ankle in the sagittal plane for the 10, 8, 6, 4, and 2mm perturbations were, 4.17 ± 0.81°, 5.02 ± 1.46°, 2.65 ± 0.83°, 2.80 ± 0.78°, and 1.85 ± 1.21°, respectively. The TBI group mean range about the ankle was 5.17 ± 1.91°, 2.94 ± 2.05°, 1.95 ± 0.26°, 1.86 ± 0.87°, and 1.48 ± 0.31°, respectively. The TBI group showed a higher mean range than the HC group for the 10mm perturbation, but their mean range was smaller than the HCs for the other 4 conditions, and this pattern was also reflected in the RMS values (Fig 2).

The mean range of the HC group knee angle in the sagittal plane for the 10, 8, 6, 4, & 2mm perturbations were 5.13 ± 2.50°, 6.67 ± 5.09°, 2.72 ± 0.99°, 2.30 ± 1.16°, and 1.44 ± 0.51° respectively, and the mean range for the TBI group knee angle in the sagittal plane for the same perturbations were 9.97 ± 5.70°, 4.85 ± 5.08°, 1.85 ± 0.60°, 2.12 ± 1.08°, and 1.48 ± 0.59° respectively. The TBI group range was larger than the HC group for the 10 & 2mm perturbations, and decreased in line with the decreasing perturbation amplitudes with the exception of the 4mm perturbation which increased. The HC group range increased for the 8mm trial, but then decreased in line with the decreasing perturbation amplitudes for the remaining trials. These patterns were mostly reflected in the mean RMS values as well, except both group's values increased for the 4mm condition, and the TBI group mean RMS was smaller than the HC group for all conditions except the 10mm perturbation, as shown in Fig. 2.

The HC group mean range about the hip in the sagittal plane for the 10, 8, 6, 4, & 2mm perturbations were 3.50 ± 1.44°, 3.32 ± 1.56°, 2.22 ± 1.30°, 2.08 ± 0.96°, and 1.31 ± 0.62° respectively. The TBI group mean range about the hip for the same perturbations were 12.97 ± 11.86°, 5.98 ± 3.14°, 3.77 ± 2.53°, 3.73 ± 1.95°, and 2.01 ± 0.50° respectively. The TBI group's mean range values were larger than the HC group's for all conditions, and both the TBI and HC group values decreased with perturbation amplitude for all conditions. The mean RMS values for the TBI group were larger than the HC group for all conditions and decreased with perturbation amplitude for all conditions, while the HC group instead increased slightly for the 8 & 4mm conditions.

B. *Statistics*

No significant outliers were observed in the data. The joint range of movement was found to be normal ($p>0.05$) for 12 out of the 15 mean range values of joint movement (dependent variable). Maulchy's test of sphericity for the platform perturbations was non-significant ($p>0.05$), but was significant for the joints ($p=0.002$). As a result, sphericity was assumed for the within-subject factor, platform perturbations, while sphericity was not assumed for the within-subject factor, joints, and the metrics from Greenhouse-Guisser are reported. The main effect of joints was not significant ($F(1.5,9)=1.615$, $p=0.247$). However, a significant joint*group interaction effect was observed ($F(1.5,9)=4.731$, $p=0.047$). The significant joint*group interaction effect shows that mean range values of joint movement had different effects for the TBI and HC groups. The main effect of platform perturbation

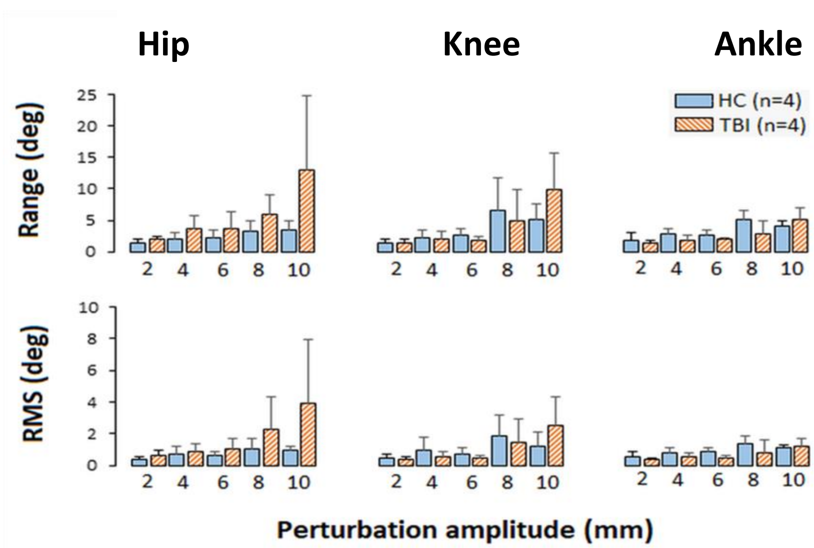

Figure 2. The mean range and RMS values for hip, knee and ankle joints across all perturbation amplitudes for the HC (n=4) and TBI (n=4) groups. The perturbation amplitudes are sorted from low to high order for intuitive presentation of the data.

was significant ($F(4,24)=9.434$, $p<0.001$), which showed that the platform perturbations were different. However the platform perturbation*group interaction effect was not significant ($F(4,24)=2.731$, $p=0.053$). Tests of between-subjects effects showed that the TBI and HC groups were not significantly different ($F(1,6)=0.719$, $p=0.429$).

IV. Discussion

The objective of this investigation was to examine the presence, absence or alteration of the fundamental postural control strategies using lower extremity joint kinematics in individuals with TBI during perturbed standing. Previous research has shown that balance dysfunction after a TBI is characterized by muscle weakness, reduced motor control and coordination, and the reduced ability to perceive the position and configuration of their body, and detect changes to their base of support [2-4]. As a result, it was hypothesized that the TBI group would show a larger range of movement about their ankles, or motion about both their hips and ankles in a combined strategy, as motion about their ankles alone would not be sufficient to compensate for the perturbation to their base of support, due to muscle weakness, or a delay in their response to the perturbation, either of which could produce the COM adjustment required to maintain stability to exceed the torque, angle or COM displacement limits of the ankle joint, thereby necessitating the invlovment of the hip joint.

The results show a substantial difference in movement about the hip between the two groups, and the TBI group had a much larger mean range and RMS than the HC group. Changes in the range and RMS values for the TBI group were much larger as the amplitude of the perturbation changed, while the HC groups inter-trial changes were much smaller. This may indicate that the TBI group was substantially more reliant on their hip joints to maintain their balance during the perturbations, particularly during the largest perturbation, and this reliance was well coupled with perturbation amplitude. This may indicate that the given perturbations were ‘too challenging’ for the TBI group to rely on the ankle strategy alone, and as a result, they had to shift to the hip strategy to maintain their balance. Conversely, the HC group’s reliance on their hips was minimal and was much less coupled to the perturbation amplitude. This difference in hip reliance by group is further reinforced by the fact that the joint*group interaction effect was found to be significant. These findings are also in agreement with the motor coordination errors reported previously [12]. During the platform-induced sway, it was observed that the TBI participant’s kinematic response was characterized by elevated hip flexion, delayed reversal of ankle motion from dorsiflexion to plantar flexion and increased net angular displacement at the knee [12]. These responses were observed for 1 Hz sinusoidal perturbations in the current study, particularly for 10mm trials (Fig. 2). However, lack of information on joint response timing and electromyography data limit the interpretation of the current investigation.

Inter-trial variability observed in the present study can be explained anecdotally. Several participants were not expecting the initial 10mm perturbation to be so substantial, and this caused them to elicit an over-compensatory response to the onset of the perturbation. For the 8mm perturbations, the participants did not appear to display such a response, as they may have over-prepared for this subsequent perturbation. It will be interesting to investigate if there were any anticipatory neuromuscular responses prior to the onset of the perturbations as reported in [13].

The ability to broadly apply the results of this study are limited by the small sample size and the limited number of trials available for analysis for each participant. Further investigation into the kinematic aspects of postural control in individuals post TBI should be performed with larger sample sizes and a larger number of trials and additional trials can be allocated to permit the participants to familiarize themselves with the perturbations to allow for more consistent results over a greater number of trials. Heterogeneity of the current sample in terms of TBI severity could also have induce the variability seen in the responses. Hence, future studies should include a more homogenous sample of TBI participants. Nonetheless, the current work is one of the few studies that demonstrate lower limb kinematic differences between HC and TBI and presents preliminary evidence of increased sagittal plane hip movement during perturbed standing in the TBI group.

References

[1] L. M. Nashner, C. L. Shupert, F. B. Horak, and F. O. Black, "Organization of posture controls: an analysis of sensory and mechanical constraints," *Prog Brain Res,* vol. 80, pp. 411-8; discussion 395-7, 1989.

[2] M. Campbell and A. Parry, "Balance disorder and traumatic brain injury: preliminary findings of a multi-factorial observational study," *Brain Inj,* vol. 19, pp. 1095-104, Dec 2005.

[3] R. Pilkar, K. K. Karunakaran, A. Veerubhotla, N. Ehrenberg, O. Ibironke, and K. J. Nolan, "Evaluating Sensory Acuity as a Marker of Balance Dysfunction After a Traumatic Brain Injury: A Psychophysical Approach," *Front Neurosci,* vol. 14, p. 836, 2020.

[4] J. J. Sosnoff, S. P. Broglio, S. Shin, and M. S. Ferrara, "Previous mild traumatic brain injury and postural-control dynamics," *J Athl Train,* vol. 46, pp. 85-91, Jan-Feb 2011.

[5] T. Paillard and F. Noe, "Techniques and Methods for Testing the Postural Function in Healthy and Pathological Subjects," *Biomed Res Int,* vol. 2015, p. 891390, 2015.

[6] D. C. Bland, C. Zampieri, and D. L. Damiano, "Effectiveness of physical therapy for improving gait and balance in individuals with traumatic brain injury: a systematic review," *Brain Inj,* vol. 25, pp. 664-79, 2011.

[7] F. B. Horak and L. M. Nashner, "Central programming of postural movements: adaptation to altered support-surface configurations," *J Neurophysiol,* vol. 55, pp. 1369-81, Jun 1986.

[8] C. F. Runge, C. L. Shupert, F. B. Horak, and F. E. Zajac, "Ankle and hip postural strategies defined by joint torques," *Gait Posture,* vol. 10, pp. 161-70, Oct 1999.

[9] L. M. Nashner and G. McCollum, "The organization of human postural movements: A formal basis and experimental synthesis," *Behavioral and Brain Sciences,* vol. 8, pp. 135-150, 1985.

[10] D. A. Winter, "Human balance and posture control during standing and walking," *Gait Posture,* vol. 3, pp. 193-214, 1995.

[11] R. B. Davis, S. Õunpuu, D. Tyburski, and J. Gage, "A gait analysis data collection and reduction technique," *Human Movement Analysis,* vol. 10, pp. 575-587, 1991.

[12] A. Shumway-Cook and R. Olmscheid, "A systems analysis of postural dyscontrol in traumatically brain-injured patients," *Journal of Head Trauma Rehabilitation,* vol. 5, pp. 51-62, 1990.

[13] R. Pilkar, O. Ibironke, N. Ehrenberg, and K. J. Nolan, "Anticipatory and Compensatory Postural Responses during Perturbed Standing in Individuals with Traumatic Brain Injury," *Annu Int Conf IEEE Eng Med Biol Soc,* vol. 2019, pp. 5080-5083, Jul 2019.