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Original Research Article

An Experiment in Healthy Volunteers Examined the Impact of Virtual Reality's Optokinetic Stimulation on Weight-Bearing Shifts While Walking.

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Abstract

Aim: Effect of Optokinetic Stimulation in The Virtual Reality Environment on Weight-Bearing Shift During Gait Movement in Healthy Subjects.

Methods: This cross-sectional study was done the Department of PMR, Patna Medical College and Hospital, Patna, Bihar, India for 1 year. after taking the approval of the protocol review committee and institutional ethics committee. 50 healthy subjects, 30 males and 20 females (22.1 ± 1.7 years), were included in this study. After applying the following exclusion criteria: impaired vision, visual field disturbances, and orthopedic disorders significantly affecting the gait of participants, 30 subjects participated in the static balance test, and 30 in the gait test; 10 subjects (7 males and 3 females) participated in both tests.

Results: During HOKS and TOKS, CFPs on both sides (right, 0.67 ± 0.34 cm; left, -0.08 ± 0.38 cm for HOKS, and right, 0.73 ± 0.48 cm; left, -0.14 ± 0.48 cm for TOKS) significantly shifted to the right side compared to those during stationary conditions; consequently, the left CFP significantly shifted to the medial side, and the right CFP to the lateral side. These results demonstrated a rightward weight-bearing shift on the foot sole surface in both feet during TOKS and HOKS, which also shows that a rightward shift in weight bearing is associated with not only right-left foot balance but also the balance within each foot.

Conclusion: OKS via HMD-VR could induce a significant weight-bearing shift, and significantly change the gait parameters. OKS via a VR environment can have potential implications for rehabilitation after stroke.

Keywords: Optokinetic stimulation Head-mounted display; Virtual reality; Center of pressure; Weight-bearing; Posture balance; Gait characteristics.

Introduction

Patients with hemiplegia after stroke have a characteristic standing posture, preferring the non-paretic side for bearing weight, inducing paretic/non-paretic asymmetry[1]Individuals bearing more weight on the non-paretic side often demonstrate greater postural sway of the center of pressure (CoP) during static standing[2]and impairment in the activities of daily living[3]and walking[4]

The extent of weight-bearing asymmetry in walking is related to the gait velocity and the step length on the paretic side[4,5]The asymmetric ratio of single-support time is directly proportional to the risk of falls[6]Therefore, reducing lateral asymmetry would improve the walking ability and prevent falls; this is significant as 51% of patients with chronic stroke experience falls while walking[7]Increasing weight-bearing on the paretic side during static standing and correcting lateral asymmetry may solve these problems[8]In the subacute phase after stroke, approximately corresponding to a period from 1 week to 3 months, patients can sit without support but cannot retain independent standing; the sway of the CoP (*i.e.*, instability of posture balance) has been reported to increase with deviation of weight-bearing to the non-paretic side[9,11]Many studies (*e.g.* Pournajaf *et al.*)[12]have investigated sitting ability in the subacute phases after stroke. Initial sitting inability and the inability to independent walk were significantly correlated with an ability to walk independently in later stages (usually later than 6 months after stroke). Duarte *et al.* analyzed the extent of sitting balance ability as a predictor of walking ability in later stages[13]Other studies have evaluated the predictive value of recovery of sit- ting balance for walking ability 6 months after stroke, or at discharge[14,15]Early rehabilitation reportedly provides good outcomes[16,17] Therefore, early improvement of the sitting posture balance by adjusting weight-bearing deviation would improve the walking ability of patients and pre- vent falls.

Although various types of sensory stimulation, *e.g.*, somatosensory, vestibular, and visual, have been shown to provide a certain effect on improving the postural deficits in patients after stroke under static standing conditions[18,20]few studies have focused on the effect of the visual approach on the unilateral CoP shift, particularly in patients who are incapable of independent standing. Many previous studies on visual approach using optokinetic stimulation (OKS) in healthy subjects have focused on either an increase in sway of the CoP, *i.e.*, postural instability, or a higher cognitive process, *i.e.*, vection[21]but not on the unilateral weight-balance shift. Regarding previous re- ports on the combined effect of visual feedback with conventional physical therapy in stroke patients, some studies consider it effective in improving posture balance, but others do not[22,24]

Material and methods

This cross-sectional study was done the Department of PMR, Patna Medical college and Hospital, Patna, Bihar, India for 1 year, after taking the approval of the protocol review committee and institutional ethics committee. 50 healthy subjects, 30 males and 20 females (22.1 ± 1.7 years), were included in this study. After applying the following exclusion criteria: impaired vision, visual field disturbances, and orthopedic disorders significantly affecting the gait of participants, 30 subjects participated in the static balance test, and 30 in the gait test; 10 subjects (7 males and 3 females) participated in both tests.

Static balance test

An immersive VR environment was projected, using astereoscopic HMD (Oculus Rift CV1, Oculus VR, LLC, Irvine, Newport Beach, CA, USA). The virtual image system was developed, using Unity3D (Unity Technologies, San

Francisco, CA, USA). Three thousand small white virtual spheres (each measuring 25 cm in diameter) were randomly distributed on the inner, black-colored surface of a virtual sphere that had a 16-m radius from the center of the subject's eyes. For OKS, the entire 3D field with the small spheres was rotated around the subject's longitudinal axis (HOKS), or frontal axis (TOKS) at one of five different velocities (i.e., 20, 40, 60, 80, and 100°/s), with each velocity being maintained for 30 s. We routinely used a rightward direction for HOKS, and a clockwise direction for TOKS. A stationary visual state that displayed the same optokinetic pattern with no movement was used as control (stationary condition).

Participants wearing the HMD stood quietly for 40 s during each recording (10 s for the waiting period, and 30 s for the recording period), with their feet placed parallel to each other on a stabilometric platform (Gravicoder GP-5000, ANIMA Co., Ltd., Tokyo, Japan) which is used to record CoP displacement.

The participants were told to look at the whole visual field. CoP data, which were collected at a sampling frequency of 100 Hz, were evaluated offline using MATLAB (Math Works Inc., Natick, MA, USA) to analyze the following CoP parameters: sway path (SP), sway area (SA), sway vector, sway mean, and sway slope. SP is the total distance of CoP displacement on a two-dimensional surface of the stabilometric platform which was obtained by combining the x-axis (right-left axis), and y-axis (antero-posterior axis) components. SA was calculated by multiplying the distribution width (distance between the highest and lowest x-axis coordinates) by the height (distance between the highest, and lowest y-axis coordinates) of the CoP trajectory during each recording period. Sway vector (SV) is the distance from the origin of the coordinates, and the counterclockwise angle from the x-axis (rightward: 0°; leftward: 180°) of the CoP position at each sampling time which was obtained based on the study of Kitabayashi et al[25]The counterclockwise angle was divided into eight sectors, and each sector had an angle of 45°. Each SV was classified into eight groups according to the sector it fell into. Moreover, sway means were the average values of CoP positions of the x-and y-axis components. Sway slopes were calculated to examine the time course of the CoP deviation. For each condition, the slope was based on least squares linear regression approximation fitted over a recording period of CoP trajectory data of x-or y-axis components.

Gait analysis

The gait test was performed in a flat unobstructed hallway with a floor-line marked at a distance of 4 m. To ensure a natural gait, the cadence, and speed of walking were not controlled. Similar to the static balance experiment, the gait test employed HOKS and TOKS, using an HMD at a velocity of 40°/s. The subjects maintained an upright position for 15 s at the beginning of the gait test and were asked to look at the whole visual field. Subsequently, they started walking at their own pace during which OKS was maintained continuously, and the trajectory was measured using simultaneous video recordings of the front and side directions. Retro-reflective markers were placed on the chest, and acromion of each subject. Kinovea motion detection software was used for the offline analysis of the two-dimensional reconstruction of the walking trajectory. The rightward deviation of the walking trajectory was quantified by fitting a quadratic function: y (t)=k•t2 , where y is the distance of the y-coordinate of the trajectory position at time (t) from the x-axis (a straight line along which the subjects are instructed to walk down), and k is the quadratic function coefficient. Foot pressure (FP) was recorded using an insole pressure recording system (BodiTrak Insole System, VISTA MEDICAL, Ltd., Winnipeg, MB, Canada). The center of the foot pressure position (CFP) was calculated according to the method of Luximon et al[26]to indicate the CoP deviation in the mediolateral direction from the foot centerline during the stance phase. Positive and negative CFP values indicated lateral and medial deviations, respectively.

Statistical analysis

Statistical analysis was performed using Statistical package for social sciences 25.0 medical software package. For the static bas used to evalualance test, a one-way analysis of variance (ANOVA) with Dunnett's multiple comparison post hoc test wate the effects of OKS velocity on the CoP parameters. A paired t test was employed to compare the SP between HOKS and TOKS for different OKS velocities. For the gait analysis, one-way ANOVA and Dunnett's multiple comparison post hoc tests were used to evaluate the differences in the values of the gait parameters among the following three conditions: stationary, HOKS, and TOKS. $p<0.05$ was considered as statistically significant. Sample size was calculated, using the G*power software, version 3.0. A result with a value of >80% was considered adequate[27]

Results

CoP in the static balance test during OKS

Velocity characteristic of OKS was firstly examined in the static balance test condition. In the stationary condition, the CoP was generally distributed in the area near the center. In a previous study, both HOKS and TOKS tended to increase displacement of CoP position, and CoP was spread over the distribution range during stationary condition. SA values during HOKS and TOKS were significantly larger than the SA values during the stationary state (Table 1). Correspondingly, the total SP was also significantly larger than that during the stationary condition (Table 1). In contrast to the instability increase during OKSs, the distribution range of CoP position tended to shift to the rightward direction based on the three different line colors, representing the three transition time periods during recording. This deviation in the two-dimensional space was evaluated using the following three metrics: SV, sway mean, and CoP slope. The SV magnitude for the example in along the 0° direction (rightward) was 0.71, 0.98, and 2.52 cm for the stationary condition, HOKS, and TOKS, respectively. The differences in the corresponding means of the SV magnitude among the subjects were highly significant, indicating that the magnitude of the effect was greater in both HOKS and TOKS than in the stationary condition (Table 1). By contrast, no significant difference in SV magnitude was observed among any stimulus conditions along the 90° direction (forward), 180° direction (leftward), or 270° direction (backward). For the sway mean values along the x-axis, a significant difference was found at 40°/s for HOKS, and 20-60°/s for TOKS (Table 1). The rightward shift of CoP position was also analyzed by estimating the linear regression slope for the x-axis CoP positions. In both HOKS and TOKS, the CoP slope for all stimulus velocities was positive (with significantly greater values 40–60°/s for HOKS, and 20 - 60°/s for TOKS than for the stationary condition) (Table 1). This reveals that, with time, both HOKS, and TOKS shifted CoP rightward. The results demonstrate that HOKS and TOKS induce a lateral (rightward) CoP shift as well as an increase in the CoP sway although a slight predominance of the effect was noted in TOKS compared to HOKS. Based on the overall results of the effect of OKS velocity, we used 40°/s for the gait experiment.

Walking trajectory and gait cycle during OKS

We investigated whether 40°/s of OKS provided by HMD could evoke a significant weight-bearing shift during walking. During the stationary condition, the walking trajectories of each subject were distributed around the x-axis, with some modest fluctuation to either side .

During HOKS, a rightward deviation was induced; nevertheless, the deviation range was similar to that of the stationary condition. During TOKS, a rightward deviation was clearly observed in all subjects, with a greater deviation than that of the stationary condition.

Because the walking trajectory of most subjects deviated in a curved manner rather than linearly, a statistical fitting of a quadratic function was performed on the time course to quantify the degree of deviation. The regression coefficient for all subjects was 0.13 ± 0.9, 0.58 ± 1.3, and 2.28 ± 1.58 during the stationary condition, HOKS, and TOKS, respectively.

A significant difference was found in the means between TOKS, and the stationary condition [F (2,59) =15.79, p<0.01]; however, no difference in the mean between HOKS, and stationary condition was observed. The results suggested that TOKS had a stronger effect on shifting the walking trajectory rightward than HOKS.

Table 1:

		Stationary	**HOKS**					**TOKS**				
OKS velocity (°/s)		0	20	40	60	80	100	20	40	60	80	100
Sway path (cm)		34.8	45.2*	48.3†	49†	47.9†	47.7	71†	72.5†	65.9†	60.8†	57.8†
		(7.6)	(17.2)	(19.9)	(18.9)	(18.8)	(22.7)	(26.6)	(31.5)	(27.7)	(25.5)	(24.2)
Sway area (cm2)		3.8	6.3	6.7*	7.2*	7.9*	9.8	13.8†	15.8†	14.2†	14.7*	11.3
		(2.2)	(5.7)	(4.4)	(6.2)	(6.2)	(17.5)	(8.8)	(14.3)	(13.5)	(18.4)	(11.1)
Sway vector	rightward	0.34	0.62	0.94†	0.76*	0.87†	0.78	1.16†	1.42†	1.32†	1.3†	0.94
	(0°)	(0.26)	(0.57)	(0.87)	(0.86)	(0.66)	(0.83)	(0.77)	(0.91)	(0.76)	(0.84)	(0.59)
	forward	0.52	0.78	0.75	0.83	0.67	0.79	0.7	1.08	0.89	1.04	0.67
	(90°)	(0.42)	(0.47)	(0.72)	(0.69)	(0.5)	(0.76)	(0.64)	(1.08)	(0.84)	(0.95)	(0.57)
	leftward	0.54	0.47	0.35	0.8	0.47	0.54	0.66	0.43	0.6	0.79	0.83
	(180°)	(0.36)	(0.38)	(0.38)	(0.64)	(0.39)	(0.68)	(0.72)	(0.5)	(0.48)	(1.4)	(1.04)
	backward	0.48	0.47	0.7	0.54	0.73	0.54	0.73	0.58	0.78	0.66	0.97
	(270°)	(0.39)	(0.4)	(0.59)	(0.47)	(0.51)	(0.57)	(0.62)	(0.73)	(0.76)	(0.81)	(1.29)
Sway mean (cm)	x-axis	-0.13	0.01	0.38†	0.17	0.16	0.18	0.52†	0.81†	0.76†	0.34	0.12
		(0.4)	(0.53)	(0.53)	(0.61)	(0.5)	(0.72)	(0.9)	(0.73)	(0.68)	(1.16)	(1.06)
	y-axis	0.07	0.28	0.04	0.18	-0.02	0.24	0.07	0.37	0.17	0.17	-0.1
		(0.49)	(0.6)	(0.65)	(0.64)	(0.59)	(0.77)	(0.73)	(0.99)	(0.66)	(0.82)	(0.82)
x-axis sway slope	slope	-0.46	0.39	1.47†	0.92*	0.91	0.84	2.07†	2.58†	2.05†	1.21	-0.14
		(1.54)	(2)	(2.04)	(2.18)	(2.97)	(2.13)	(2.47)	(2.38)	(1.91)	(3.15)	(3.07)
	Pearson’s r	0.29	0.32	0.26	0.32	0.42	0.31	0.34	0.32	0.32	0.33	0.33
		(0.2)	(0.25)	(0.25)	(0.22)	(0.21)	(0.23)	(0.18)	(0.24)	(0.25)	(0.27)	(0.26)

For the gait cycle, the right leg stance phase was significantly longer, and the left leg stance phase was shorter in HOKS and TOKS than in the swing phase, magnitude relationship between in the stationary condition. By contrast, the limbs were reversed, i.e., a significantly shorter stance phase on the right leg and longer on the left leg were observed in both HOKS and TOKS. No significant difference was found in the stationary condition. These results suggest that HOKS and TOKS changed the gait cycles, thereby increasing the weight balance toward the right side by lengthening the stance phase of the right leg.

Mean foot sole pressure during stance phase

The lateral (rightward) shift found in the walking trajectory and gait cycle begs the question of whether such change would be accompanied by an actual change in the FP during the stance phase. To clarify this, mean FP during OKS was measured for both feet. A significant difference in the mean of the right FP during the stance phase between HOKS and the stationary condition was found. During TOKS, the mean of the right FP was significantly greater than the mean of the left FP, suggesting that weight bearing increased in the right foot During stationary conditions, the CFP position was located laterally from the midline of the foot sole in both feet, with a value of 0.47 ± 0.23 cm and -0.28± 0.26 cm for the right and left foot, respectively. No significant difference between the left and right foot was found (p=0.07).

During HOKS and TOKS, CFPs on both sides (right, 0.67 ± 0.34 cm; left, -0.08 ± 0.38 cm for HOKS, and right, 0.73 ± 0.48 cm; left, -0.14 ± 0.48 cm for TOKS) significantly shifted to the right side compared to those during stationary conditions; consequently, the left CFP significantly shifted to the medial side, and the right CFP to the lateral side. These results demonstrated a rightward weight-bearing shift on the foot sole surface in both feet during TOKS and HOKS, which also shows that a rightward shift in weight bearing is associated with not only right-left foot balance but also the balance within each foot.

Discussion

The purpose of our study was to clarify the effect of simple OKS in an immersive VR environment on weight-bearing shift during gait movement as well as during quiet standing. The device used in this study (Oculus Rift) could provide not only a whole-field visual scene motion but also a high-fidelity VR environment through a positional-tracking sensor of the wearer's head motions that adjust the external image according to the head motion. This device, which was first developed for academic research, could also be used in designs, business, arts, and entertainment, and its effectiveness is demonstrated in various fields, including medical education, and museum exhibition description[28,29]Our results showed that OKS via HMD-VR induces a significant increase in both stance time and FP on the stimulation side, which in turn resulted in a lateral deviation of the walking direction. These findings strongly suggest that a weight-bearing shift toward the stimulation direction during gait and that the device could serve as a useful exercise tool during gait for patients with stroke.

Patients with hemiplegic stroke with a higher degree of lateral asymmetry have an increased risk of falls[30,31]Previous reports suggested that rehabilitation, using task-related training, or treadmill training combined with video games, or a visual scene motion on a large screen could improve such asymmetry compared to the standard strengthening methods with traditional gymnasium equipment[32,33]In our study, we showed that the use of wearable HMD-VR device during gait may benefit not only the inpatients but also those with stroke who were discharged home.

The OKS in a VR environment in this study contained 3100 small virtual spheres distributed randomly at 15-m distance from the subject's eyes. In our preliminary study, the optimal set of parameters, such as the number and size of the small virtual spheres, and distance (radius of the whole virtual space sphere) were determined based on the results of the magnitude of lateral CoP sway, or occasionally, illusory effect perceived by several subjects. For OKS velocity, 40°/s of OKS was routinely used for both HOKS and TOKS to induce deviation to a lateral direction based on the overall effect on the CoP sway parameters. Moreover, at a velocity >40°/s, approximately one fifth of the subjects (HOKS, 16.67% (5/30); 20% (6/30), TOKS) experienced some dizziness. In the gait test, no subject experienced either dizziness, or discomfort with the velocity. Nevertheless, visual display technologies, especially HMD, have generally been shown to be associated with "cybersickness", resulting in nausea, headaches, and dizziness[34,35]The velocity of 40°/s seems to be a moderate condition that is suitable for weight-bearing shift and reduces the subject's load, which could be an important consideration for clinical use. The OKS velocities used in previous studies were as follows: HOKS and/or TOKS, 20°/s[36,37]TOKS, 40°/s; HOKS or TOKS, 60°/s[38]and HOKS, 20–100°/s. Although the OKS in previous studies were presented by a pattern rotation or screen projection, the range of velocities seems to be similar to that used in our study. However, in the previous studies, the effect of CoP deviation to a certain direction has not been consistently observed. This could be because the focus of the reports was on the overall increase in CoP sway (i.e., instability). To the best of our knowledge, no study has focused primarily on the effect of OKS via immersive HMD-VR on weight-bearing shift, especially during gait movement from the viewpoint of rehabilitation.

In this study, we did not test our stimulus condition in elderly patients or those with stroke because our primary goal was to determine how to shift weight bearing in a stable and safe way. The OKS approach via HMD-VR may result in qualitatively different responsiveness among elderly patients, or those with stroke although our preliminary work with elderly subjects suggests that the preferred OKS velocity of 40°/s could produce similar weight-bearing shifts during both static and gait conditions without signs of dizziness or falls. Moreover, our study utilized an extremely simple 3D pattern. Owing to recent developments, VR technology has successfully advanced in its creation of a realistic environment by combining non-visual sensory information such as vestibular, auditory, and somatosensory cues Thus, a more realistic stimulus pattern may result in a more effective exercise training program for posture–balance recovery in patients with stroke in the near future.

Conclusion

OKS via HMD-VR could induce a significant weight-bearing shift, and significantly change the gait parameters. OKS via a VR environment can have potential implications for rehabilitation after stroke.

Reference

1. Chaudhuri S, Aruin AS. The effect of shoe lifts on static and dynamic postural control in individuals with hemiparesis. Arch Phys Med Rehabil 2000; 81:1498–503.
2. Marigold DS, Eng JJ. The relationship of asymmetric weight- bearing with postural sway and visual reliance in stroke. Gait Posture 2006; 23:249–55.
3. Sackley CM. The relationships between weight bearing asymmeti y after stroke, motor function and activities of daily living. Physiother Theory Pract 1990; 6:179–85.
4. Nardone A, Godi M, Grasso M, Guglielmetti S, Schieppati M. Stabilometry is a predictor of gait performance in chronic hemiparetic

stroke patients. Gait Posture 2009; 30:5–10.
5. Szopa A, Domagalska-Szopa M, Lasek-Bal A, Żak A. The link be- tween weight shift asymmetry and gait disturbances in chronic hemipa- retic stroke patients. Clin Interv Aging 2017; 12:2055–62.
6. Wei TS, Liu PT, Chang LW, Liu SY. Gait asymmetry, ankle spastic- ity, and depression as independent predictors of falls in ambulatory stroke patients. PLoS One 2017;12:e0177136.
7. Forster A, Young J. Incidence and consequences of falls due to stroke: a systematic inquiry. BMJ 1995; 311:83–6.
8. Geurts AC, de Haart M, van Nes IJ, Duysens J. A review of standing balance recovery from stroke. Gait Posture 2005; 22:267–81.
9. Perlmutter S, Lin F, Makhsous M. Quantitative analysis of static sitting posture in chronic stroke. Gait Posture 2010; 32:53–6.
10. Taylor D, Ashburn A, Ward CD. Asymmetrical trunk posture, uni- lateral neglect and motor performance following stroke. Clin Rehabil 1994; 8:48–53.
11. Mudie MH, Winzeler-Mercay U, Radwan S, Lee L. Training symme- try of weight distribution after stroke: a randomized controlled pilot study comparing task-related reach, Bobath and feedback training approaches. Clin Rehabil 2002; 16:582–92.
12. Pournajaf S, Goffredo M, Agosti M, Massucci M, Ferro S, Franceschini M, *et al.* Community ambulation of stroke survivors at 6 months follow-up: an observational study on sociodemographic and sub-acute clinical indica- tors. Eur J Phys Rehabil Med 2019; 55:433–41.
13. Duarte E, Marco E, Muniesa JM, Belmonte R, Diaz P, Tejero M, *et al.* Trunk control test as a functional predictor in stroke patients. J Rehabil Med 2002; 34:267–72.
14. Feigin L, Sharon B, Czaczkes B, Rosin AJ. Sitting equilibrium 2 weeks after a stroke can predict the walking ability after 6 months. Ger- ontology 1996; 42:348–53.
15. Nitz J, Gage A. Post stroke recovery of balanced sitting and ambula- tion ability. Aust J Physiother 1995; 41:263–7.
16. Musicco M, Emberti L, Nappi G, Caltagirone C; Italian Multicenter Study on Outcomes of Rehabilitation of Neurological Patients. Early and long-term outcome of rehabilitation in stroke patients: the role of patient characteristics, time of initiation, and duration of interventions. Arch Phys Med Rehabil 2003; 84:551–8
17. Matsui H, Hashimoto H, Horiguchi H, Yasunaga H, Matsuda S. An exploration of the association between very early rehabilitation and out- come for the patients with acute ischaemic stroke in Japan: a nationwide retrospective cohort survey. BMC Health Serv Res 2010; 10:213.
18. Pérennou DA, Leblond C, Amblard B, Micallef JP, Hérisson C, Pélis- sier JY. Transcutaneous electric nerve stimulation reduces neglect-related postural instability after stroke. Arch Phys Med Rehabil 2001; 82:440–8.
19. Rode G, Tiliket C, Charlopain P, Boisson D. Postural asymmetry reduction by vestibular caloric stimulation in left hemiparetic patients. Scand J Rehabil Med 1998; 30:9–14.
20. Hung JW, Chou CX, Hsieh YW, Wu WC, Yu MY, Chen PC, *et al.* Randomized comparison trial of balance training by using exergaming and conventional weight-shift therapy in patients with chronic stroke. Arch Phys Med Rehabil 2014; 95:1629–37.
21. Brandt T, Paulus W, Straube A. Vision and posture. In: Bles W, Brandt T, editors. Disorders of posture and gait. Amsterdam: Elsevier; 1986. p. 157-75.
22. Blanks RH, Fowler CG, Zizz CA, Williams KE. Postural adjustments produced by moving visual (horizontal

optokinetic) patterns. J Am Acad Audiol 1996; 7:39–48.
23. Sasaki O, Usami S, Gagey PM, Martinerie J, Le Van Quyen M, Ar- ranz P. Role of visual input in nonlinear postural control system. Exp Brain Res 2002; 147:1–7.
24. Pavlou M, Quinn C, Murray K, Spyridakou C, Faldon M, Bronstein AM. The effect of repeated visual motion stimuli on visual dependence and postural control in normal subjects. Gait Posture 2011; 33:113–8
25. Kitabayashi T, Demura S, Noda M. Examination of the factor structure of center of foot pressure movement and cross-validity. J Physiol Anthropol Appl Human Sci. 2003; 22:265-272.
26. Luximon Y, Cong Y, Luximon A, Zhang M. Effects of heel base size, walking speed, and slope angle on center of pressure trajectory and plantar pressure when wearing high-heeled shoes. Hum Mov Sci. 2015; 41:309-319.
27. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. Behav Res Methods. 2007; 39:175-191.
28. Chang KE, Chang CT, Hou HT, Sung YT, Chao HL, Lee CM. Development and behavioral pattern analysis of a mobile guide system with augmented reality for painting appreciation instruction in an art museum. Comput educ. 2014; 71:185-197.
29. Sugiura A, Kitama T, Toyoura M, Mao X. The use of augmented reality technology in medical specimen museum tours. Anat Sci Educ. 2019; 12:561-571.
30. De Quervain IA, Simon SR, Leurgans S, Pease WS, McAllister D. Gait pattern in the early recovery period after stroke. J Bone Joint Surg Am. 1996; 78:1506-1514.
31. Balasubramanian CK, Bowden MG, Neptune R, Kautz SA. Relationship between step length asymmetry and walking performance in subjects with chronic hemiparesis. Arch Phys Med Rehabil. 2007; 88:43-49.
32. Keshner EA, Kenyon RV, Langston J. Postural responses exhibit multisensory dependencies with discordant visual and support surface motion. J Vestib Res. 2004; 14:307-319.
33. Ilona JM, Ingrid GL, Meijer JG. Effect of virtual reality training on balance and gait ability in patients with stroke: systematic review and meta-analysis. Phys Ther. 2016; 96:1905-1918.
34. Howarth PA, Costello PJ. The occurrence of virtual simulation sickness symptoms when an HMD was used as a personal viewing system. Displays. 1997; 18:107-116.
35. Rebenitsch L, Owen C. Review on cybersickness in applications and visual displays. Virtual Reality. 2016; 20:101-125.
36. Chiarovano E, de Waele C, Mac Dougall HG, Rogers SJ, Burgess AM, Curthoys LS. Maintaining balance when looking at a virtual reality three-dimensional display of a field of moving dots or at a virtual reality scene. Front Neurol. 2015; 6:1-9.
37. Yelnik AP, Kassouha A, Bonan IV, Leman MC, Jacq C, Vicaut E, et al. Postural visual dependence after recent stroke: Assessment by optokinetic stimulation. Gait Posture. 2006; 24:262-269.
38. Provost CP, Tasseel-Ponche S, Lozeron P, Piccinini G, Quintaine V, Arnulf B, et al. Standing postural reaction to visual and proprioceptive stimulation in chronic acquired demyelinating polyneuropathy. J Rehabil Med. 2018; 50:278-284