Tracking gaze while walking on a treadmill: spatial accuracy and limits of use of a stationary remote eye-tracker

V. Serchi- *IEEE Student*, A. Peruzzi, A. Cereatti, and U. Della Croce

Abstract— **Inaccurate visual sampling and foot placement may lead to unsafe walking. Virtual environments, challenging obstacle negotiation, may be used to investigate the relationship between the point of gaze and stepping accuracy. A measurement of the point of gaze during walking can be obtained using a remote eye-tracker. The assessment of its performance and limits of applicability is essential to define the areas of interest in a virtual environment and to collect information for the analysis of the visual strategy. The current study aims at characterizing a gaze eye-tracker in static and dynamic conditions. Three different conditions were analyzed: a) looking at a single stimulus during selected head movements, b) looking at multiple stimuli distributed on the screen from different distances, c) looking at multiple stimuli distributed on the screen while walking. The eye-tracker was able to measure the point of gaze during the head motion along medio-lateral and vertical directions consistently with the device specifications, while the tracking during the head motion along the anterior-posterior direction resulted to be lower than the device specifications. During head rotation around the vertical direction, the error of the point of gaze was lower than 23 mm. The best accuracy (10 mm) was achieved, consistently to the device specifications, in the static condition performed at 650 mm from the eye-tracker, while point of gaze data were lost while getting closer to the eye-tracker. In general, the accuracy and precision of the point of gaze did not show to be related to the stimulus position. During fast walking (1.1 m/s), the eye-tracker did not lose any data, since the head range of motion was always within the ranges of trackability. The values of accuracy and precision during walking were similar to those resulting from static conditions. These values will be considered in the definition of the size and shape of the areas of interest in the virtual environment.**

I. INTRODUCTION

In the elderly population, falling represents a common problem. Approximately 30 % of older adults experience at least a fall every year [1]. Many of these falls occur during walking [2] and gait variability represents a predictor of fall risk [3], [4]. Walking safely requires both accurate foot placement and visual sampling [5], which refers to the extraction of visual information from the environment to be used in path planning, especially in challenging footpaths [6]. Visual impairment reduces the ability to detect hazards in the environment, and represents an additional important predictor of fall risk [7], [8].

Research supported by the region of Sardinia (Italy)

V. Serchi, A. Peruzzi, A. Cereatti and U. Della Croce are with POLCOMING Department, Information Engineering Unit, University of Sassari, Sassari, Italy and with the Interuniversity Centre of Bioengineering of the Human Neuromusculoskeletal System (e-mail: vserchi@uniss.it).

Visual sampling during walking is affected by age [9], [10], oculomotor and locomotor deficits [11], terrain conformity [12] and constraints [13]. Specifically, unsatisfactory visual sampling strategies have been observed in elderly fallers when asked to step into a target [5], [13]. Recently, gait rehabilitation programs, based on the combined use of virtual reality and treadmill, aimed at enhancing gait [14]–[17] and at reducing fall risk [16], [18]–[20] in pathological populations. These studies required the participants to negotiate virtual obstacles [14], [15], [17], [18], [20] and the evaluation focused more on the motor performance than the visual strategy. Since an accurate foot placement requires both intact visual and motor abilities [21], virtual environments, including obstacles, may be used to investigate the relationship between the point of gaze and stepping accuracy. The combined use of virtual reality and treadmill may provide safe, highly controlled and custom experimental environments [22]. The analysis of the point of gaze measurements requires the identification of areas of interest (AoIs), defined as the regions included in the visual stimulus (on the screen) from which information is extracted. In defining the contour of AoIs, the knowledge of the accuracy and precision of the eye-tracker measurements is essential [23].

The current study aims at characterizing a commercial eye-tracker [24] used for the estimation of the point of gaze in static and dynamic conditions. The point of gaze was measured with a remote eye-tracker to limit the discomfort during movement. Remote eye gaze trackers are usually less accurate than wearable ones in the presence of head and body movements, however they are easier to setup and more comfortable [25].

The accuracy and precision of the eye-tracker measurements were assessed in three different conditions: a) looking at a single stimulus during selected head movements, b) looking at multiple stimuli distributed on the screen at various distances and c) looking at multiple stimuli distributed on the screen while walking.

II. MATERIALS AND METHODS

A. Experimental setup

A remote eye-tracker (Tobii, TX300, sampling frequency: 300 Hz) was used to measure the point of gaze of a healthy male participant (age: 48 y.o., height: 1.87 m). According to the manufacturer specifications, the eye-tracker allows to collect data as long as the head motion is within a certain range (±150 mm along the anterior-posterior –AP– direction and ±100 mm along the medio-lateral –ML– and vertical –V– directions). The eye-tracker was placed on an adjustable tripod, about 1.5 m above the ground, and in front of a treadmill. A monitor (47-inch LCD, 1280x1024 px), used to display visual stimuli, was attached to the wall in front of the treadmill. The eye-tracker and the monitor were centered with respect to the treadmill. While the subject was standing on the treadmill, the eye-tracker was inclined to center the subject's eyes (Fig. 1). A six-camera stereo-photogrammetric system (Vicon, T20, sampling frequency: 100 Hz) was used to track the subject's head. Four markers were attached on a headband and four markers were placed on the eye-tracker. The subject's point of gaze was calibrated using a nine-point calibration grid provided by the eye-tracker software [26]. The distance between the subject's eyes and the eye-tracker was set to 650 mm, according to the device guidelines. During the calibration, the subject was asked to look at the points presented at nine different known positions on the screen. The results provided by the calibration were within the level of precision and accuracy provided by the manufacturer [24]. The subject was required to focus his gaze on a visual stimulus, consisting in two concentric black and white circles in a black background. The acquisitions were carried out in a darkened room. The starting distance between the subject's eyes and the center of the eye-tracker was 650 mm. Three different conditions were tested.

1) A single visual stimulus was located at the center of the screen (Fig. 1) and was displayed for about 60 s. The subject was asked to focus on the stimulus, while performing four different tasks:

- standing while translating along AP axis (*tAP*),
- standing while translating along ML axis (*tML*),
- standing while translating along V axis (*tV*),
- standing while rotating the head around V axis (*rV*).

2) A visual stimulus was displayed in 25 consecutive positions on the screen. The stimulus persisted in each position for two seconds. The first nine positions were those of the calibration grid (P1:P9). The 16 remaining positions (PA:PR) were located according to Fig. 2.

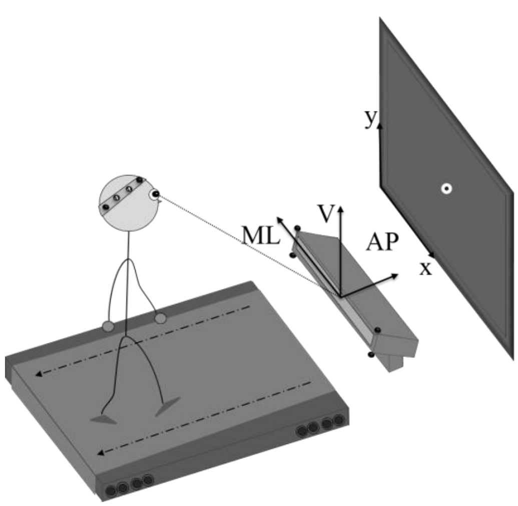

Figure 1. Experimental set-up and markers placement.

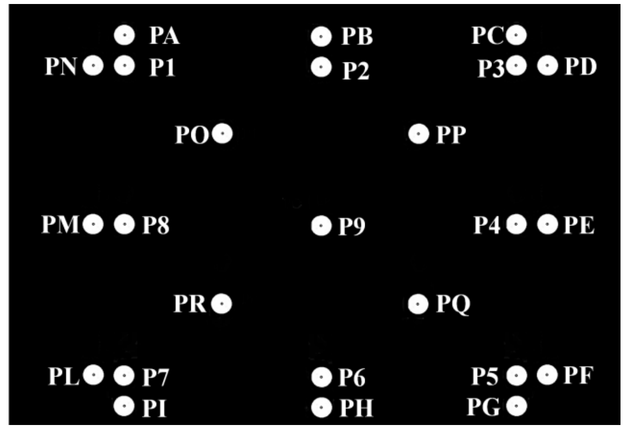

Figure 2. The 25 points-grid used to display the optical stimuli in the second experimental condition.

The subject was standing at a) 550 mm (*st550*), b) 650 mm (*st650*) and c) 750 mm (*st750*) from the eye-tracker.

3) The subject walked on the treadmill at 1.1 m/s, while focusing his gaze on the 25 consecutive positions of the visual stimulus (*walk*).

The eye-tracker and the stereo-photogrammetric systems were synchronized to acquire simultaneously. Two trials were recorded for each task.

B. Data processing

Two separate filters were used to remove eyes blinks, short gaze deviations and flickering from point of gaze data [27]. The head range of motion (RoM) and the range of trackability (RoT - minimum and maximum value of head motion during gaze tracking) were estimated for the first and the third conditions.

For the first condition, for each sample i (i=1,...,N) the eye-tracker measured the x- and y-coordinates of the point of gaze (PoG_i). The monocular (${}^{L}e_i$ and ${}^{R}e_i$) and the binocular errors (e_i) were computed as the differences between the PoG_i and the stimulus position. Also, the maximum binocular error value e_{MAX} was computed.

For the second and the third conditions, for each stimulus position (j, j=1,..,25), the PoG, was averaged over the N samples. The corresponding bias b_j and standard deviation sd_j were calculated. The average bias and standard deviation values, computed over the 25 stimulus positions, were also calculated together with their minimum and maximum values. The eye-tracker measurements, obtained for each stimulus position in the two repetitions, were averaged.

III. Results

Results relative to the first condition are reported in Table I. During *tAP*, *tV*, *rV* the subject's eyes were partially lost during the task execution (RoT<RoM). Conversely, during *tML*, the subject's eyes were never lost within the tested RoM. The error values for a repetition of the task *rV* are depicted in Fig. 3. When this task is performed, the left eye is lost while rotating the head to the left and the right eye is lost

while rotating the head to the right. Moreover, the monocular errors increased when increasing rotation angles of the head.

Average, minimum and maximum values for bias and standard deviation in correspondence of *st550*, *st650*, *st750* and *walk* tasks are reported in Table II and Table III, respectively.

The *PoG* for a subset of the 25 stimulus positions is shown in Fig. 4 for the second and the third conditions together with a representation of its dispersion (*sd*<5 mm small radius circle, 5 mm<*sd*<10 mm medium radius circle, *sd*>10 mm large radius circle). One of the two trials of the task *st550* had no valid PoG_i in correspondence of stimulus position P3. None of the trials *st550* presented valid PoG_i in correspondence of stimulus position P4 (missing circle in Fig. 4).

In Table IV, the RoM of the head for the *walk* task is reported. The RoT is not reported since during the task, the subject's eyes were never lost.

IV. Discussion

In the present study, the characterization of a remote eye gaze tracker in static and dynamic conditions was performed. In the first condition, the performance of the eye-tracker was investigated during several head movements. The analysis allowed for the definition of the limits within which a subject is free to move his/her head, while the eye-tracker tracks his/her point of gaze. The measured RoT along ML and V directions are consistent with those provided by the manufacturer, while the measured RoT along the AP direction is reduced (±80 mm instead that ±150 mm).

TABLE I. The RoM of the head, the RoT and the e_{MAX} for tasks in the first condition

Task	RoM		RoT		e_{MAX}
	min	*max*	*min*	*max*	
tAP [mm]	540	740	576	740[a]	30[b]
tML [mm]	-135	105	-135[a]	105[a]	29[b]
tV [mm]	-110	63	-90	63[a]	28[b]
rV [deg]	-61	62	-48	55	23[b]

a. The point of gaze was correctly measured without interruption, and, therefore, the corresponding maximum or minimum values of the RoM have been reported.

b. [mm].

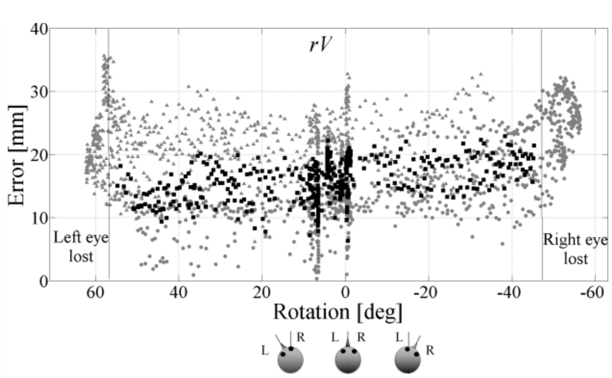

Figure 3. Point of gaze error estimation vs head rotation around the vertical axis. In the figure $^{L}e_i$ is reported as grey triangles, $^{R}e_i$ as grey dots and the e_i as black squares.

TABLE II. The average, the minimum and the maximum values of the bias of the *PoG* for the tasks in the second and third conditions

[mm]	*st550*	*st650*	*st750*	*walk*
avg(*b*)	18	10	19	14
max(*b*)	76	23	39	32
min(*b*)	6	1	2	1

TABLE III. The average, the minimum and the maximum values of the standard deviation of the *PoG* for the tasks in the second and third conditions

[mm]	*st550*	*st650*	*st750*	*walk*
avg(*sd*)	4	4	6	6
max(*sd*)	12	10	11	20
min(*sd*)	1	2	2	2

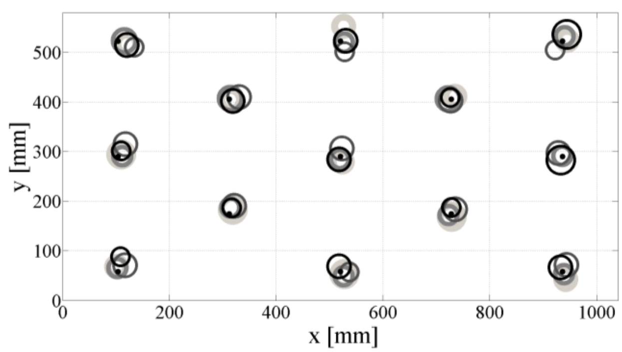

Figure 4. The point of gaze measurements for a subset of the 25 stimulus positions (black dots) for the second and third conditions. The size of circles is representative of the dispersion around their mean value (circle center): thick light grey: *st550*, thick dark grey: *st650*, thin grey: *st750*, thin black: *walk* (1) small radius circle *sd*<5 mm; 2) medium radius circle 5 mm<*sd*<10 mm; 3) large radius circle *sd*>10 mm). Therefore the size of the circles in the figure is a qualitative description of the actual dispersion of the PoG. No circle is reported if no PoG was recorded.

TABLE IV. The RoM of the head along AP, ML, V and around V, during the walk task

Head motion	RoM	
	min	*max*
AP translation [mm]	654	770
ML translation [mm]	-98	-10
V translation [mm]	-50	6
V rotation [deg]	-9	19

As expected, the error values in *rV* are lower when the head rotation is within the range ±20 deg. However, for this task, the error values were lower than 23 mm within the RoT.

During the second and third conditions, the entire screen area has been characterized in terms of accuracy (bias) and precision (standard deviation). In general, the accuracy and the precision of the point of gaze measurements did not show any remarkable dependency on the stimulus position. The best accuracy was achieved, consistently to the manufacturer specifications, in *st650* (*b*=10 mm). Conversely, the precision of the point of gaze seems not to be influenced by the eyes-device distance.

The most critical task was *st550*, which presented no signal for position P4 (Fig. 4). This is not surprising, since it is consistent with the results of the first condition, where the tracking was lost during the approaching phase of the translation along the AP direction (RoT min=576 mm).

During the third condition, the eye-tracker measured the point of gaze during fast walking without losing data, since the range of motion of the head was always within the range of trackability. The *walk* task, compared to the static tasks in the second condition, showed similar results in terms of accuracy and precision of the point of gaze.

The preliminary results of this study provide insights for the design of an experimental protocol involving gaze recordings carried out with a remote eye-tracker while the subject moves. Such protocols may be effectively used in virtual reality based applications, in which the analysis of gaze is under investigation. In fact, in those applications requiring the use of AoIs, their number, size, shape and distance should be selected taking into consideration the results of this study to avoid false positive determinations of AoI. Therefore, based on [23] recommendations, AoIs minimal size should be decided in according to the maximum value of bias achieved during walking (30 mm). Moreover, considering the maximum value of the standard deviation achieved during walking, a margin of at least 20 mm around this region should be added.

REFERENCES

[1] A. J. Blake, K. Morgan, M. J. Bendall, H. Dallosso, S. B. Ebrahim, T. H. Arie, P. H. Fentem, and E. J. Bassey, “Falls by elderly people at home: prevalence and associated factors,” *Age Ageing*, vol. 17, pp. 365–372, 1988.

[1] M. Sartini, M. L. Cristina, A. M. Spagnolo, P. Cremonesi, C. Costaguta, F. Monacelli, J. Garau, and P. Odetti, “The epidemiology of domestic injurious falls in a community dwelling elderly population: an outgrowing economic burden,” *The European Journal of Public Health*, vol. 20, no. 5, pp. 604–606, 2009.

[2] J. M. Hausdorff, D. A. Rios, and H. K. Edelberg, “Gait variability and fall risk in community-living older adults: A 1-year prospective study,” *Arch. Phys. Med. Rehabil.*, vol. 82, pp. 1050–1056, 2001.

[3] J. Verghese, R. Holtzer, R. B. Lipton, and C. Wang, “Quantitative gait markers and incident fall risk in older adults,” *J. Gerontol. A. Biol. Sci. Med. Sci.*, vol. 64, no. 8, pp. 896–901, Aug. 2009.

[4] G. J. Chapman and M. A. Hollands, “Evidence that older adult fallers prioritise the planning of future stepping actions over the accurate execution of ongoing steps during complex locomotor tasks,” *Gait Posture*, vol. 26, no. 1, pp. 59–67, Jun. 2007.

[5] A. E. Patla, “Visual Control of Human Locomotion,” *Advances in Psychology*, vol. Volume 78, 1991, pp. 55–97.

[6] R. J. Reed-Jones, S. Dorgo, M. Hitchings, and J. Bader, “Vision and agility training in community dwelling older adults: Incorporating visual training into programs for fall prevention,” *Gait Posture*, vol. 35, no. 4, pp. 585–589, 2012.

[7] M. C. Nevitt, S. R. Cummings, S. Kidd, and D. Black, “Risk factors for recurrent nonsyncopal falls. A prospective study,” *JAMA*, vol. 261, no. 18, pp. 2663–8, May 1989.

[8] W. R. Young and M. A. Hollands, “Evidence for age-related decline in visuomotor function and reactive stepping adjustments,” *Gait Posture*, vol. 36, no. 3, pp. 477–481, 2012.

[9] G. J. Chapman and M. A. Hollands, “Age-related differences in stepping performance during step cycle-related removal of vision,” *Exp. brain Res.*, vol. 174, no. 4, pp. 613–21, Oct. 2006.

[10] K. A. Crowdy, M. A. Hollands, I. T. Ferguson, and D. E. Marple-Horvat, “Evidence for interactive locomotor and oculomotor deficits in cerebellar patients during visually guided stepping,” *Exp. Brain Res.*, vol. 135, no. 4, pp. 437–454, Dec. 2000.

[11] D. S. Marigold and A. E. Patla, “Gaze fixation patterns for negotiating complex ground terrain,” *Neuroscience*, vol. 144, no. 1, pp. 302–13, Jan. 2007.

[12] G. J. Chapman and M. A. Hollands, “Age-related differences in visual sampling requirements during adaptive locomotion,” *Exp. Brain Res.*, vol. 201, no. 3, pp. 467–78, Mar. 2010.

[13] A. Peruzzi, A. Cereatti, A. Mirelman, and U. Della Croce, “Feasibility and Acceptance of a Virtual Reality System for Gait Training of Individuals with Multiple Sclerosis,” *Eur. Int. J. Sci. Technol.*, vol. 2, no. 6, pp. 171–181, 2013.

[14] Y.-R. Yang, M.-P. Tsai, T.-Y. Chuang, W.-H. Sung, and R.-Y. Wang, “Virtual reality-based training improves community ambulation in individuals with stroke: a randomized controlled trial,” *Gait Posture*, vol. 28, no. 2, pp. 201–6, Aug. 2008.

[15] A. Mirelman, I. Maidan, T. Herman, J. E. Deutsch, N. Giladi, and J. M. Hausdorff, “Virtual reality for gait training: can it induce motor learning to enhance complex walking and reduce fall risk in patients with Parkinson’s disease?,” *J. Gerontol. A. Biol. Sci. Med. Sci.*, vol. 66, no. 2, pp. 234–40, Feb. 2011.

[16] D. L. Jaffe, D. A. Brown, C. D. Pierson-Carey, E. L. Buckley, and H. L. Lew, “Stepping over obstacles to improve walking in individuals with poststroke hemiplegia,” *J. Rehabil. Res. Dev.*, vol. 41, no. 3A, p. 283, May 2004.

[17] A. Mirelman, L. Rochester, M. Reelick, F. Nieuwhof, E. Pelosin, G. Abbruzzese, K. Dockx, A. Nieuwboer, and J. M. Hausdorff, “V-TIME: a treadmill training program augmented by virtual reality to decrease fall risk in older adults: study design of a randomized controlled trial,” *BMC Neurol.*, vol. 13, no. 1, p. 15, Feb. 2013.

[18] M. W. van Ooijen, M. Roerdink, M. Trekop, J. Visschedijk, T. W. Janssen, and P. J. Beek, “Functional gait rehabilitation in elderly people following a fall-related hip fracture using a treadmill with visual context: design of a randomized controlled trial,” *BMC Geriatr.*, vol. 13, p. 34, Jan. 2013.

[19] G. Paolini, A. Peruzzi, A. Mirelman, A. Cereatti, S. Gaukrodger, J. Hausdorff, and U. Della Croce, “Validation of a Method for Real Time Foot Position and Orientation Tracking With Microsoft Kinect Technology for Use in Virtual Reality and Treadmill Based Gait Training Programs,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 1, no. 1, 2013.

[20] R. F. Reynolds and B. L. Day, “Visual guidance of the human foot during a step,” *J. Physiol.*, vol. 569, pp. 677–684, 2005.

[21] M. K. Holden, “Virtual Environments for Motor Rehabilitation : Review,” *Cyberpsychology Behav.*, vol. 8, no. 3, pp. 187-211, 2005.

[22] K. Holmqvist, M. Nyström, and R. Andersson, Eye tracking: A comprehensive guide to methods and measures. OXFORD University, 2011, pp. 1–702.

[23] J. D. Morgante, R. Zolfaghari, and S. P. Johnson, “A Critical Test of Temporal and Spatial Accuracy of the Tobii T60XL Eye Tracker,” *Infancy*, vol. 17, no. 1, pp. 9–32, 2012.

[24] C. H. Morimoto and M. R. M. Mimica, “Eye gaze tracking techniques for interactive applications,” *Comput. Vis. Image Underst.*, vol. 98, no. 1, pp. 4–24, Apr. 2005.

[25] User Manual — Tobii Studio, Ver 3.2. Tobii Tecnology AB, 2012.

[26] G. Veneri, F. Rosini, P. Federighi, A. Federico, and A. Rufa, “Evaluating gaze control on a multi-target sequencing task: the distribution of fixations is evidence of exploration optimisation,” Comput. *Biol. Med.*, vol. 42, no. 2, pp. 235–44, Feb. 2012.