2010

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Dimitrios Kastavelis
University of Nebraska at Omaha

Mukul Mukherjee
University of Nebraska at Omaha, mmukherjee@unomaha.edu

Leslie M. Decker
University of Nebraska at Omaha

Nicholas Stergiou
University of Nebraska at Omaha, nstergiou@unomaha.edu

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Dimitrios Katsavelis, *University of Nebraska, Omaha NE*
Mukul Mukherjee, *University of Nebraska, Omaha NE*
Leslie Decker, *University of Nebraska, Omaha NE*
Nicholas Stergiou[1], *University of Nebraska Medical Center, Omaha NE*

Abstract: *Backward walking (BW) shows significant differences with forward walking (FW) and these differences are potentially useful in rehabilitation. However the lack of visual cues makes BW risky. The purpose of this study was to investigate the effect of visual cues provided by a virtual environment on FW and BW on gait variability. Each subject underwent four conditions of treadmill walking at self-selected pace. The subjects walked backwards in three conditions and forwards in the fourth condition. A virtual corridor was displayed to the subjects in the FW condition (forward optic flow) and two of the backward conditions (forward and backward optic flow). The third BW condition was a control condition (no visual cues). Gait variability measures of the hip, knee and ankle range of motion and the stride interval were analyzed. Magnitude of variability was evaluated with the coefficient of variation and structure of variability with approximate entropy. Significant differences were demonstrated between the FW and the BW gait characteristics as well as in gait variability (for both magnitude and structure of variability). No significant differences were found between the three BW conditions as a result of the direction of visual cues. In order to get optimal benefit of BW in the aged and the diseased, optical flow of visual feedback may need to be manipulated in a different manner than FW. Future studies will explore other parameters of visual cues like the velocity of optic flow and appearance of obstacles to obtain the best visual cue configuration for rehabilitation.*

Key Words: virtual reality, nonlinear analysis, locomotion, vision, optic flow.

INTRODUCTION

Backward walking (BW) shows characteristic gait patterns. Some of these patterns like the movement of the hip angle during the gait cycle are time-reversed mirror images of the forward walking (FW) gait cycle (Thortensson, 1986; Winter, Pluck & Young, 1989). However, these same studies also

[1] Correspondence address, Dr. Nicholas Stergiou, Nebraska Biomechanics Core Facility, University of Nebraska at Omaha, Omaha, Nebraska 68182-0216. E-mail, nstergiou@mail.unomaha.edu .

demonstrated that the knee and the ankle angle patterns during the gait cycle were different between BW and FW patterns. Moreover, in terms of joint torques BW and FW are highly correlated for the hip and ankle but not the knee. Furthermore, electromyographic (EMG) studies of FW and BW locomotor patterns have shown ambiguous results. On one hand, there was evidence that muscle activity during FW and BW was similar and could be produced by simply reversing the temporal cycling of muscle contraction (Winter et al., 1989). On the other hand, major differences between the muscle activities have also been demonstrated during FW and BW patterns (Thortensson, 1986; Grasso et al., 1998). Greater level of EMG activity has been shown to occur during BW in comparison to FW (Grasso et al., 1998; Winter et al., 1989).

The higher physiological stress resulting from backward locomotion is advantageous for both fitness training and rehabilitation. There are distinct advantages of BW over FW. During BW both heart rate and oxygen consumption are higher than FW for the same speed (Flynn, Connery, Smutok, Zeballos, & Weisman, 1994). Similar differences are also observed for backward running in comparison to forward running. In addition, BW may provide a greater benefit for certain conditions like overuse injuries in the lower extremities and patellofemoral dysfunctions (Flynn & Soutas-Little, 1995). This occurs because during backward locomotion, patellofemoral joint reaction forces and eccentric loading of the patellar tendon are both reduced. Specifically, peak patellofemoral joint compressive forces are significantly lower and occur significantly later in the stance phase in backward locomotion in comparison to forward locomotion (Flynn & Soutas-Little, 1995). In a randomized controlled study with stroke subjects, participants who underwent backward walking training in addition to their conventional exercise regimen showed greater improvement than control group participants who received only the conventional exercise regimen (Yang, Yen, Yeng, & Lieu, 2005). In that study, significant improvements were noted in gait speed, stride length and gait symmetry. However, before BW is used for rehabilitation purposes it should be noted that specific populations like the elderly may have difficulty in adapting to BW (Laufer, 2005). Such difficulties in BW locomotor abilities may be reduced by incorporating visual cues during the training regimen.

Visual cues during locomotion impact higher brain centers. The medial superior temporal region has been implicated in the processing of optic flow (Smith, Wally, Williams, & Singh, 2006). Several studies have implicated the posterior parietal cortex in being involved in the sensorimotor integration of optic flow perception and its impact on movement performance (Bremmer, Schlack, Duhamel, Graf, & Fink, 2001; Zhang & Britten, 2004). In terms of optic flow moving towards and away from the observer, significantly different brain correlates have been demonstrated in the visual areas (Wunderlich et al., 2002).

The manipulation of visual cues during locomotion has been shown to impact the variability of the locomotor pattern (Prokop, Schubert, & Berger, 1997; Hollman, Bray, Robman, Bang, & Kaufman, 2006). In these studies, the variability measures were restricted to linear measures like standard deviation and coefficient of variation which at best can only provide information on the amount of variability and not its organization over time, i.e., structure (Sosnoff, Jordan, & Newell, 2005; Herman, Gilardi, Gurvich, & Hausdorff, 2005). Nonlinear measures like Approximate Entropy provide such answers and reveal predictability or regularity of the time series (Newell, 1997; Stergiou, Buzzi, Kurz, & Heidel, 2004).

In several studies, the locomotor patterns of FW have been shown to be affected by visual cues in both healthy subjects (Prokop et al., 1997; Verraine, Bonnard, & Pailhaus, 2002) and in patients with stroke (Lamontagne, Fung, McFadyen, & Faubert, 2007). In addition, the variability of the neuromuscular system was also shown to be affected (Hollman et al., 2006). However it is not clear how such visual cues would impact the characteristics of BW. Specifically do visual cues cause changes in BW gait in comparison to FW? Secondly, do visual cues affect variability of gait characteristics in comparison to FW? Finally, does the direction of OF affect the gait characteristics of BW? In order to answer these questions, healthy human subjects walked on a treadmill at a Self Selected Pace (SSP) with visual cues being provided by a virtual reality (VR) environment.

METHOD

Subjects

Six healthy adults (4 males, 2 females) participated in the study (age, 27.7 ±2 yr; height, 175.3 ±10 cm; weight, 68.7 ±11 kg). Subjects were free from any musculoskeletal problems and had no recent or remote history of significant lower extremity injuries that might have affected their gait. In addition, subjects were excluded from the study in case of any type of visual or vestibular deficiency. Prior to testing, each subject signed an informed consent approved by our Medical Center Institutional Review Board.

Instrumentation

The custom VR environment was written in C++ by using open graphics library (OpenGL) and graphics library utility toolkit. The immerse environment was projected by a commercial projection system (NEC Display Solutions, Itasca, IL) on a 80-inch flat screen that was positioned 3 meters away from the plane of motion. For the VR conditions, an endless virtual corridor with realistic side walls was projected onto the screen to create the VR environment, which was visible only with specialized stereoscopic glasses (Fig. 1). The motion of the projected environment was set to alter at a frequency that was matched with the speed of the treadmill. The VR environment consisted of two separate images on the screen. On viewing them through red-blue stereo glasses that the subjects wore throughout the experiment, the two images merged into a single scene providing the subject with a feeling of depth of the rendered scene.

Fig. 1. Experimental set up consisting of the treadmill, the BWS, the VR and the eight-camera motion capture system (only six are shown in the figure). The projector was located right behind and above the subject's head and three meters away from the screen to ensure a wide field of view for the subject. On the right side, the virtual reality (VR) environment is shown as it appeared to the subject without the red-blue stereo glasses. The use of a stereo graphics card (nVidia Corporation, Santa Clara, CA) renders not one but two separate images on the screen. Viewing them with special glasses creates a feeling of depth of the rendered scene.

A Motion Analysis (Motion Analysis Corp, Santa Rosa, CA) camera system was used to capture kinematics at 60 Hz while subjects walked on a motorized treadmill (312-C, Bodyguard, Canada; Fig. 1). Eight optoelectric cameras were positioned around the treadmill to collect three-dimensional trajectory data from the markers that were placed on the subjects prior to data collection. Reflective markers were placed on specific anatomical landmarks to track the motion of the hip, knee and ankle joints (Nigg, Cole, & Nachbauer, 1993). To ensure safety, each subject wore a safety vest suspended overhead throughout the course of the experiment (Fig. 1).

Experimental Design

Each subject underwent three conditions of BW and one condition of FW, with each condition being eight minutes long. Prior to the experiment, each subject was asked to walk backward on a motorized treadmill at a comfortable pace that could be easily maintained for a long time. When the subject informed the investigator that such a speed was identified, the value was recorded and was used subsequently for the BW conditions as their self-selected pace (SSP). A similar protocol was followed to obtain the SSP for FW. Therefore the three BW conditions were performed at the SSP for the BW and the single FW condition was performed at the SSP for the FW. By using a self-selected pace, any variability changes detected were due to the independent variable and not due to probable discomfort that may be associated with using a pre-determined speed for all subjects (Jordan, Challis, & Newell, 2007). Following the determination of the SSP, subjects were asked to walk backwards on the treadmill for at least six minutes as a warm up/familiarization period. This familiarization period is considered sufficient for the achievement of reliable measurements (Matsas, Taylor, & McBurney, 2000).

Gait variability measurements are speed-dependent (Jordan et al., 2007). In other words, variations in walking velocity affect the magnitude of variability present in the locomotion patterns. Thus, the application of a constant-driven treadmill can facilitate the measurement of movement variability by controlling for speed variations in comparison to overground walking where such measurements would therefore become difficult. During testing, subjects walked on a treadmill at their SSP under four different conditions, (1) BW with no optic flow ($BACK_{nVR}$), (2) BW with OF perceptually equivalent to the SSP ($BACK_{OFb}$), (3) BW with OF perceptually equivalent to the SSP, but in the opposite direction ($BACK_{OFf}$) and (4) FW with OF perceptually equivalent to the SSP ($FORW_{OFf}$). The four conditions were presented in random order. Subjects had the option to rest between conditions.

Data Analysis

Eight minutes of continuous unfiltered data were analyzed so as to get a more accurate representation of the variations within the system (Rapp, 1994; Mees & Judd, 1993). Furthermore, since the same instrumentation was used for all subjects, it was assumed that the level of measurement noise would be consistent for all subjects and that any differences could be attributed to changes within the system itself. Therefore, filtering the data may have eliminated important information and provided a skewed view of the system’s inherent variability.

The unfiltered time series of the marker position data in three dimensions were acquired by EVART software (Motion Analysis Corp., Santa Rosa, CA). The three dimensional angular displacements of the hip, knee and ankle joints were calculated using laboratory software developed in Matlab (Mathworks Inc., MA, USA) and according to the algorithms described by Vaughn, Davis, and O’Connor (1999). Only the sagittal angular displacement was examined because data from the other planes collected via skin markers are associated with increased error (Capozzo, Leardini, Benedetti, & Della Croce, 1996). However, we collected three-dimensional data instead of two-dimensional to increase accuracy by minimizing perspective error. After identifying the minimum and maximum joint angles for each gait cycle and for each condition, the range of motion (ROM) was calculated by subtracting the maximum and minimum values for each gait cycle. Joint kinematic variability was examined in addition to variability of the stride interval, because it has been shown that variability of joint kinematics offers a more sensitive measure of differences between groups than the variability of the stride characteristics (Barrett, Noordegraf, &Morrison, 2008). However, it has also been shown that the distribution of the stride interval may be a fractal process (Goldberger, Peng, & Lipsitz, 2002; Hausdorff et al., 1995; Hausdorf, Peng, Ladin, Wei, & Goldberger1996; West & Griffin, 1999). Therefore, variability of joint kinematics was analyzed in addition to variability of the stride interval.

Stride interval was defined as the time duration between two consecutive maximum angular positions of the knee joint. All variables were unidimensional, since they were measured once per gait cycle. Subsequently, means and the coefficient of variations (CV) were calculated from 350 consecutive gait cycles for each dependent variable in each testing condition and from each subject. It should be mentioned that eight minutes of continuous BW produced on average 415 gait cycles for each condition. However, the final number of gait cycles was truncated to 350 because this was the least number of gait cycles performed in any of the tested conditions by a subject. Therefore time series data corresponding to the 350 gait cycles were used for comparisons between subjects. This number is considered adequate for the nonlinear analysis performed in this study (Stergiou et al., 2004).

In addition to analyzing the magnitude of variability in this study, the structure of variability was also explored (Sosnoff et al., 2005; Stergiou et al., 2004). The structure of variability was investigated using the nonlinear method of Approximate Entropy (ApEn) which is a measure of quantifying the predictability or regularity of a time series (Pincus & Goldberger, 1994; Ryan, Goldberger, Pincus, Mietus, & Lipsitz, 1994). A time series that is predictable and regular is also less complex. If there is a change in complexity of the time series, it may indicate reorganization of the available degrees of freedom (Newell, 1997; Vaillancourt & Newell, 2000). ApEn is a measure of the logarithmic probability that a series of data points a certain distance apart exhibit similar relative characteristics on the next incremental comparison within the state space (Pincus & Goldberger, 1994). A time series with similar distances between data points results in lower ApEn values, while large differences in distances between data points results in higher ApEn values. The ApEn algorithm was implemented in MatLab where all time series were analyzed (with m, the number of observation windows to be compared = 2 and r, the tolerance factor = 0.2). All the three measures used – means, CV and ApEn were calculated for the ROM of each joint and for the stride interval.

Statistical Analysis

Group means for all dependent variables were calculated for each condition. A three way repeated measure ANOVA was performed using SPSS (14.0, Chicago, IL, USA) to determine significant overall effects of the following factors, VR condition (4 levels, $BACK_{nVR}$, $BACK_{OFb}$ $BACK_{OFf}$ and $FORW_{OFf}$), type of measurement (3 levels, Mean, CV, and ApEn) and biomechanical variable (4 levels, ROM of the hip, knee and ankle joints and stride interval). Pairwise comparisons with Bonferroni adjustment were performed to determine specific differences between the four VR conditions. However since very different types of measurements (mean and variability, linear and nonlinear variables) formed part of the analysis, separate repeated measures ANOVAs was performed on the group means for each of the four biomechanical variables (ROM of the hip, knee and ankle joints and stride interval) and for each of the three measurements (Mean, CV, and ApEn) to determine the effect of the repeated factor – VR condition (4 levels, $BACK_{nVR}$, $BACK_{OFb}$ $BACK_{OFf}$ and $FORW_{OFf}$). Post-hoc analysis using Bonferroni adjustments were performed to identify significant differences among groups. The level of significance was set at $\alpha = 0.05$.

RESULTS

The three-way repeated measures ANOVA revealed an overall effect of the VR condition ($F_{6,30} = 64.327$, $p = 0.000$). This means that the different VR conditions brought about a significant change in the biomechanical variables tested. There was a significant interaction between the VR condition and the type of measurement ($F_{3,15} = 69.012$, $p = 0.000$). This means that the effect of the VR conditions on the biomechanical variables were different across the measurements. The interaction between the VR condition with type of biomechanical variable – ROM of the hip, knee, ankle joint and stride interval was also significant ($F_{9,45} = 6.591$, $p = 0.000$). This means that the effect of the VR conditions was different across the biomechanical variables. To determine the specific locations of the differences in the VR conditions, separate repeated measures ANOVAs followed by post-hoc analysis (using Bonferroni corrections) were carried out for each type of biomechanical variable (ROM of the hip, knee, ankle joint and stride interval) across each of the three types of measurements (Mean, CV, and ApEn). The significant differences are shown in Figs. 2 and 3.

For the effect of visual cues on FW and BW gait characteristics, significant differences in mean values were obtained between the FW and BW conditions at the knee ($F_{3,15} = 21.178$, $p = 0.000$) and at the hip ($F_{3,15} = 25.964$, $p = 0.000$). For the knee and the hip ROM, the $FORW_{OFf}$ condition elicited higher mean values than all three BW conditions. The ankle ROM and the stride interval did not reveal significant differences.

For the effect of visual cues on gait variability during FW and BW, significant differences in the magnitude of gait variability were obtained between the CV of FW and BW conditions for the ankle ROM ($F_{3,15} = 11.662$, $p = 0.000$), the knee ROM ($F_{3,15} = 13.655$, $p = 0.000$), the hip ROM ($F_{3,15} = 18.728$, $p = 0.000$), and the stride interval ($F_{3,15} = 14.247$, $p = 0.000$). For each of the biomechanical variables, the $FORW_{OFf}$ condition elicited lower mean CV values than all three BW conditions.

Regarding the structure of gait variability, significant differences were obtained between the ApEn of FW and BW conditions for the ankle ROM ($F_{3,15} = 89.050$, $p = 0.000$), the knee ROM ($F_{3,15} = 153.138$, $p = 0.000$), the hip ROM ($F_{3,15} = 420.458$, $p = 0.000$). The stride interval did not reveal significant differences ($F_{3,15} = 2.486$, $p = 0.1$). For each of the biomechanical variables including the stride interval, the $FORW_{OFf}$ condition elicited higher mean ApEn

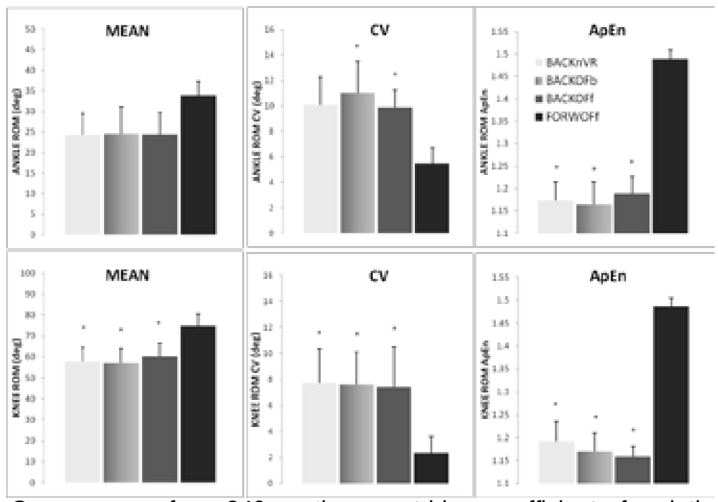

Fig. 2. Group means from 349 continuous strides, coefficient of variation (CV), and Approximate Entropy (ApEn) values for the ROM of the ankle and the knee at each condition ($BACK_{nVR}$ = backward walking without a VR environment; $BACK_{OFb}$ = backward walking in VR environment with backward optical flow; $BACK_{OFf}$ = backward walking in VR environment with forward optical flow, and $FORW_{OFf}$ = forward walking in VR with forward optic flow). * indicates significant differences in comparison to $FORW_{OFf}$ at $p < 0.05$.

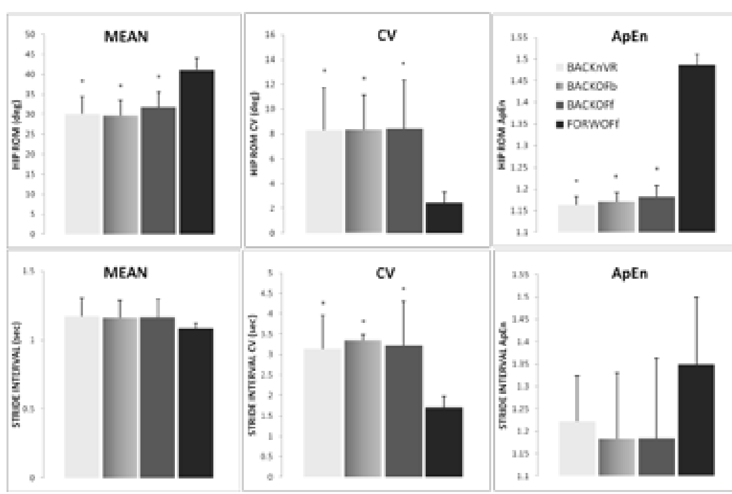

Fig. 3. Group means from 349 continuous strides, coefficient of variation (CV), and Approximate Entropy (ApEn) values for the ROM of the hip joint and for stride interval at each condition ($BACK_{nVR}$ = backward walking without a VR environment; $BACK_{OFb}$ = backward walking in VR environment with backward optical flow; $BACK_{OFf}$ = backward walking in VR environment with forward optical flow, and $FORW_{OFf}$ = forward walking in VR with forward optic flow). * indicates significant differences in comparison to $FORW_{OFf}$ at $p<0.05$.

values than all three BW conditions.

Regarding the structure of gait variability, significant differences were obtained between the ApEn of FW and BW conditions for the ankle ROM ($F_{3,15}$ = 89.050, p = 0.000), the knee ROM ($F_{3,15}$ = 153.138, p = 0.000), the hip ROM ($F_{3,15}$ = 420.458, p = 0.000). The stride interval did not reveal significant differences ($F_{3,15}$ = 2.486, p = 0.1). For each of the biomechanical variables including the stride interval, the $FORW_{OFf}$ condition elicited higher mean ApEn values than all three BW conditions.

For the effect of direction of OF on BW, no significant differences were obtained regarding comparisons between the three BW conditions.

DISCUSSION

The aims of this study were the following, first, to determine if there were differences in the mean gait characteristics due to visual cues between FW and BW, second, to determine if there were differences in gait variability due to visual cues during FW and BW and third, to determine if changing the direction of OF affected the gait characteristics of BW.

The Effect of Visual Cues on FW and BW Gait Characteristics

Regarding the mean values of the three joint ROMs – hip, knee and ankle and stride interval, significant differences were obtained between the FW and BW conditions for all variables except the stride interval. The mean ROMs of the three joints for the FW and the BW are comparable to the findings of Winter and colleagues (1989). Several studies have demonstrated that FW and BW have different biomechanical characteristics (Thortensson, 1986; Winter et al., 1989). These studies showed that the BW and FW differ in terms of gait cycle patterns as well as joint torques albeit some very characteristic time-reversed features like hip angular motion during the gait cycle. Electromyographic studies of FW and BW locomotor patterns have shown major differences between the muscle activities during FW and BW gait patterns (Grasso et al., 1998; Thortensson, 1986; Winter et al., 1989). These studies have shown a greater level of EMG activity during BW in comparison to FW. The cause of the lower mean ROM of the three joints during BW could be due to the greater control imposed on these joints demonstrated by increased muscle activity. Therefore in accordance with our findings, FW and BW may not differ in terms of a simple reversal.

The Effect of Visual Cues on Gait Variability during FW and BW

Both measures (CV and ApEn) demonstrated significant differences in variability between FW and BW. This is the first time that gait variability has been studied during BW. The magnitude of gait variability was quantified using CV and demonstrated significant differences between FW and BW for each of the three joint ROMs and stride intervals. The CV values demonstrated much higher values for the three BW conditions than the FW condition. As mentioned before, past work has shown that EMG activity during BW is greater than that during FW (Grasso et al., 1998; Thortensson, 1986; Winter et al., 1989). This increase in activity of the muscles might have led to an increase in the magnitude of variability observed at the joint. It has been observed in past literature that the magnitude of variability in muscle activity is associated with the magnitude of muscle activity in a phenomenon known as signal dependent noise (Jones et al., 2002). To our knowledge, this is also the first time that a nonlinear measure of variability have been used to identify significant effects on the structure of the gait variability as a result of VR although the importance of such a measure in normal and pathological locomotion has been revealed in the past (Dingwell & Casumano, 2000; Hausdorff et al., 1996; Slifkin & Newell, 2000; Stergiou et al., 2004). The study revealed the unique abilities of the nonlinear measure to extract pertinent information from the same data set as a linear measure. For example the comparison between the CV for $FORW_{OFf}$ and $BACK_{nVR}$ was not significant at the ankle although the ApEn values for the same comparison were different. This could mean that ApEn was a more sensitive measure of variability than CV or that the comparison differed in the structure but not in magnitude of variability.

The ApEn was found to be a highly sensitive measure of variability. The measure ApEn, quantifies randomness in a system thereby providing insight into the underlying system organization (Pincus & Goldberger, 1994). Reduction in randomness, indicated by reduction in ApEn values, is demonstrated by systems that are moving towards relative constriction (Newell, 1998). In the present study, the values in the FW condition were significantly higher. Considering that BW is an inherently more difficult/demanding task than FW, it can be considered to be comprised of fewer degrees of freedom than the less novel and more natural FW condition. Reduction in ApEn in the BW conditions indicates “freezing” of the degrees of freedom to increase stability in a more difficult and relatively unknown skill. A lower ApEn value as a sign of instability has been shown to exist in Parkinson’s patients (Vaillancourt & Newell, 2000; Vaillancourt, Sifkin & Newell, 2001), in abnormal physiology (Fleisher, 1993; Pincus, 2000; Veldhuis & Pincus, 1998) and also during the normal aging process (Newell, 1997). In fact, during normal aging, both the very young and the elderly have low ApEn values while the values for the adult were found to be higher (Newell, 1997). In a remarkable study with collegiate football athletes who suffered concussion, it was shown that in comparison to a preseason healthy status, both linear (equilibrium scores) and nonlinear (approximate entropy of the center of pressure time series) measures were affected less than 48 hours after the concussion (Cavanaugh et al., 2006). However, between 48 to 96 hours after the concussion, although the equilibrium score returned to normal, the ApEn values stayed affected. This is very important considering that it has a tremendous impact on return-to-play decisions. In the recent study, a lower ApEn can be considered a sign of greater instability and this result for BW can be correlated to past studies giving evidence for BW being a physiologically more demanding task than FW.

In terms of motor control of BW, it has been suggested that it is under the control of separate neural networks than those for FW and that these two separate networks do not interfere with each other (Choi & Bastian, 2007). We have added to this burgeoning research field by demonstrating for the first time, to our knowledge, that differences in variability also exist between the two types of locomotion on exposure to visual cues. Although the manipulation of the visual environment affects the perception of optic flow (Durgin & Gigone, 2007) and also has been shown to affect locomotion (Prokop et al., 1997; Mulavara et al., 2005), the results demonstrated in this study were largely due to the differences in the direction of walking. Although distinct brain correlates have been implicated for the perception of forward and backward optic flow (Wunderlich et al., 2002), no significant kinematic differences were found when the direction of OF was manipulated in the BW conditions. The reasons are discussed in the following paragraph.

The Effect of Direction of OF on BW

There may be several reasons for the lack of significant findings. First, in order to perform nonlinear analysis of the structure of variability, the duration of each trial was set to be eight minutes. This inadvertently might have lead to adaptive mechanisms coming into play, removing the effect optic flow on BW. Indeed, with enough adaptation, even characteristics of BW have been shown to become comparable to FW (Ung, Imbeault, Ethier, Brissi, & Capaday, 2005). Backward optic flow may be giving the correct feedback of self perception of motion during BW in comparison to the non-VR and forward optic flow conditions. However, these differences were so small that significance is lost when the novelty and difficulty of the BW task is considered. This brings us to the second reason for the lack of significant differences. Backward locomotion is a novel task for most people and is also physiologically more demanding. In comparison to forward locomotion, heart rate and oxygen consumption have been shown to be higher during backward locomotion (Hooper et al., 2004; Flynn et al., 1994). In the study by Hooper and colleagues (2004), it was shown that BW, in comparison to FW, had significantly higher percentage of maximum heart rate and also the percentage of maximum oxygen consumption (both values were about 24% higher for BW). In addition a greater level of EMG activity has also been shown during BW in comparison to FW gait (Grasso et al., 1998; Winter et al., 1989) giving more evidence to the increased effort required for performing the task. Therefore, given the greater cardiopulmonary and muscular effort required to perform the BW task, subtle changes in the direction of optical flow would not produce significant differences and more sensitive measures may be required to bring out these differences.

A limitation of the study is that a control condition with the subject walking forward and optical flow moving backwards has not been incorporated. This condition would have allowed us to compare the effect of an equivalent false visual feedback to the $BACK_{OFb}$ condition. In addition, although past literature is clear that BW is more demanding than the FW task, a measure of the subject's perception and stress levels during the BW conditions would have demonstrated that the changes observed were due to BW also being perceived as a more difficult task. In a series of ongoing studies we are incorporating these additional variables and measures.

In conclusion, the investigation of the effects of visual cues on forward and backward treadmill walking revealed that these two types of gait patterns differ in variability and this difference is not only present in the magnitude of variability but also in its structure. In order to get optimal benefit of backward locomotion in the aged and the diseased, optical flow of visual feedback may need to be manipulated in a different manner than FW. However further work is required to determine the critical parameters of such visual feedback during BW before it can be incorporated into rehabilitation regimens.

ACKNOWLEDGMENTS

This research was funded by grants awarded to Dr. Stergiou by the Nebraska Research Initiative and the NIH (K25HD047194); to Dimitrios Katsavelis by the University of Nebraska Medical Center (doctoral fellowship) and to Dr. Mukul Mukherjee by the American Heart Association (#0820136Z).

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