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THE INFLUENCE OF VISUAL PERCEPTION OF SELF-MOTION ON LOCOMOTOR ADAPTATION TO UNILATERAL LIMB LOADING

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Running Head: Virtual Reality and Unilateral Limb Loading

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ABSTRACT

Self-perception of motion through visual stimulation may be important for adapting to locomotor conditions. Unilateral limb loading is a locomotor condition that can improve stability and reduce abnormal limb movement. In this study we investigated the effect of self-perception of motion through Virtual Reality (VR) on adaptation to unilateral limb loading. Healthy young adults, assigned to either a VR or a Non-VR group, walked on a treadmill in the following three locomotor task periods – no load, loaded and load removed. Subjects in the VR group viewed a virtual corridor during treadmill walking. Exposure to VR reduced cadence and muscle activity. During the loaded period, the swing time of the unloaded limb showed a larger increase in the VR group. When load was removed, the swing time of the previously loaded limb and the stance time of the previously unloaded limb showed larger decrease and the swing time of the previously unloaded limb showed a smaller increase in the VR group. Lack of visual cues may cause the adoption of cautious strategies (higher muscle activity, shorter and more frequent steps, changes in the swing and stance times) when faced with situations which require adaptations. VR technology, providing such perceptual cues, has an important role in enhancing locomotor adaptation.

Keywords: virtual reality, vision, gait, motor learning, ankle weight.

INTRODUCTION

The ability to walk may seem to be one of the most trivial and undemanding activities of daily living. However it encompasses a highly complex ability to adapt to alterations in the environmental demands and task constraints. These adaptive changes can be classified temporally into two stages – first, immediate responses and second, long term and slower adaptive changes. The latter are crucial for learning adequate and appropriate responses to novel environmental and/or task constraints (Lam et al., 2006; Morton and Bastian, 2006; Reisman et al., 2007; Richards et al., 2007; Gordon et al., 1995). This highly complex ability involves the integration of visual, proprioceptive and vestibular sensory information to sense movement errors and correct for them. However, when sensory information is available from multiple sources and they are at conflict, visual information is given precedence (Bagesteiro et al., 2006; Flanagan and Rao, 1995; Wolpert et al., 1994; Wolpert et al., 1995). Visual sensory information may therefore play a key role in learning or re-learning locomotor dynamics.

In our daily life we perform diverse locomotor tasks in varied environments. The performance of such tasks incorporates the accurate control of the dynamics of the subject's limbs and the interacting environment. Sensorimotor abnormalities make such control of dynamics difficult to achieve. Through rehabilitation, the ability of the patient to meet task and environmental demands are improved which lead to an improvement in the quality of life. Age-related and disease-related asymmetrical gait patterns (Plotnik et al., 2007; Olney et al., 1994; Hsu et al., 2003; Lamontagne and Fung, 2004) may benefit by training to walk under different task constraints for each leg. This may be achieved in various ways: a) by using a split-belt treadmill (Reisman et al., 2007; Choi and Bastian, 2007), b) a robotic device that is attached to the lower limbs and can provide resistance during treadmill walking (Lam et al., 2006), c) actuators that provide assistance or resistance to the swing leg during treadmill training (Kurz and Stergiou, 2007), and d) by unilateral limb loading with ankle weights that provide resistance to the swinging leg (Byrne et al., 2002).

Unilateral limb loading is a type of adaptive activity that causes changes in spatio-temporal gait patterns (Skinner and Barrack, 1990, Byrne et al., 2002) as well as physiological changes (Claremont and Hall, 1988; Miller and Stamford, 1987). Specifically, limb loading causes changes in swing times, stance times and in the coordination pattern between the limb segments. It was shown that the loaded limb had increased swing phase and reduced stance phase and the unloaded limb had reduced swing phase and increased stance phase (Smith and Martin, 2007). When the load is removed there is a reversal of the effects i.e., the loaded limb had reduced swing phase and increased stance phase and the unloaded limb had increased swing phase and reduced stance phase. Moreover, changes in muscle activity have also been shown during loading and unloading tasks in healthy humans (Stephens and Yang, 1999; Bachmann et al., 2008). In these studies it was shown that during loading, muscle activities tend to increase and during unloading, they decrease. In a recent study, unilateral ankle-foot loading was shown to cause increased activity in the hip musculature and increased hip moments (Gordon et al., 2009). It has been suggested that the CNS has the ability to control timing and amplitude of muscular activity during limb loading (Stephens and Yang, 1999) however it is not clear whether external sensory feedback has the ability to influence this control.

Unilateral limb loading has been used as a therapeutic tool to reduce abnormal limb movement and improve stability (Hewer et al., 1972; Morgan, 1975). It has been used in the past to simulate paretic gait in normal subjects (Eke-Okoro et al., 1985) and in a recent paper, unilateral limb loading was tested in stroke subjects as a method to stabilize the affected leg during aquatic treadmill walking (Jung et al., 2010). Although unilateral limb loading may have clinical efficacy, walking on a treadmill with a load attached on one leg is not an easy task especially for the elderly or pathological populations like those with stroke. However, this task may be made easier if optic flow was provided to the subjects. It is known that optic flow providing perception of self motion affects several spatio-temporal patterns during treadmill walking (Prokop et al., 1997; Verraine et al., 2002; Katsavelis et al., 2010a and b). However, it is not clear if such perception of self-motion can also influence treadmill walking with unilateral limb loading. Moreover, during locomotor rehabilitation of patient populations which usually occur in static environments like walking on a treadmill in a room, visual feedback (like the perception of self motion) may not be easy to provide. Virtual Reality (VR) technology that provides perception of self motion may be the answer to this problem.

Virtual reality acts as a way of providing self perception of motion to the individual during walking. Such perception of self motion can be influenced by manipulating the optical flow of the simulated environment (Rieser et al, 1995). Simply manipulating the visual scene's velocity[1] was shown to affect the preferred walking speed (Mohler et al., 2007) and the segmental kinematics during steady standing (Dokka et al., 2009). The present research study investigated the impact of visual perception of self motion on these changes in healthy young adults. Specifically, the aim of the study was to determine whether the addition of visual perception of motion while adapting to unilateral limb loading affected gait patterns and muscle activity. The visual perception of self motion was provided through a Virtual Reality (VR) environment. All subjects adapted to unilateral loading with an ankle weight while walking on a treadmill at a self selected pace. The Optic Flow (OF) of the VR environment was synchronized to the speed of the treadmill. We expected that in the absence of VR (for control subjects), an accurate estimation of

[1] The optic flow referred here is the forward optic flow.

the speed of the treadmill would be difficult with unilateral limb loading and subjects would tend to demonstrate changes in gait kinematics and muscle activity characterizing overestimation of the treadmill velocity (intuitively underestimation would be unsafe) – higher cadence and larger muscle activity. However we also expected the subjects to be adapted to the unilateral limb loading over time. Therefore, we hypothesized that exposure to the virtual environment would produce short-term changes in loading and unloading as well as long-term adaptive changes in temporal gait patterns and muscle activity in comparison to the condition where VR was absent. Specifically, the utilization of visual cues through VR, would reduce cadence, reduce muscle activity and increase the magnitudes of changes in swing and stance times that occur during over ground walking with ankle loading.

METHODOLOGY

Subjects: Sixteen healthy adults (7 males and 9 females) participated in the study (Age: 28.13±3.67 yrs, Height: 171.25±9.50 cm, Weight: 158.75±22.17 lbs)[2]. 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 the testing the subjects signed an informed consent approved by the University's Medical Center Institutional Review Board.

Instrumentation: A custom VR environment, written in C++ using the open graphics library (OpenGL; Silicon Graphics Inc., Sunnyvale, CA) was created and was projected by a commercial projection system (Optoma TX 774, Optoma Technology Inc., Milpitas, CA) on a 98" X 68" inch flat screen (Stewart Filmscreen Corp., Torrance, CA) that was positioned 1.5 meters away from the plane of motion. This created a vertical and horizontal field of view of 59.39 and 79.42 degrees respectively. For the VR condition, an endless virtual corridor with realistic side walls was projected onto the screen to create the VR environment (figure 1). The virtual walls of the VR environment extended about 10 meters in front of the subject and did not extend behind the subject. The velocity of motion of the projected environment was synchronized to match the treadmill speed. As part of the experimental design (described later in the experimental design section), the subject underwent a treadmill walking trial during which the subject determined the preferred walking speed (the subject's average and not the instantaneous speed), by using the treadmill controls manually. This speed was noted by the experimenter and manually entered into the VR program using a keyboard function. This input was the speed at which the projected environment moved. This was how the speed of the OF in the VR environment was synchronized to the treadmill speed. Thus, the OF of the virtual

[2] Means ± standard deviation

corridor was perceptually equivalent to the speed of the treadmill. The VR environment consisted of two separate images projected on the screen. By 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. It is important for subjects to report the perception of self-motion and the sense of presence (Jancke et al., 2009) which refers to their feeling of being immersed in the virtual environment while being unaware of their real location and the technology that produces the VR environment (Wirth et al., 2007). In this particular study no subjective measures of immersion in the virtual environment were noted. However the same virtual environment and equipment that was used in this study has been used in several other virtual reality studies in our laboratory (Katsavelis et al., 2010a and b). The "Presence" questionnaire (Witmer and Singer, 1998) was used as a subjective measure of immersion in these studies. Both in terms of involvement/immersion of the subject in the VR environment and the effect of the environment on the subjects' senses, a rating close to 75% was obtained. A rating of 100% is for a VR environment of the highest order causing complete immersion of the subject.

(INSERT FIGURE 1 HERE)

Weight: an ankle weight equivalent to 10lb was attached around the left leg of each subject during the adaptation part of the experiment. Based on past literature, this was considered sufficient to perturb gait and cause locomotor adaptation (Byrne et al., 2002; Skinner and Barrack, 1990; Miller and Stamford, 1987). *Stride Analyzer:* A commercial footswitch system – stride analyser (B & L Engineering, Tustin, CA) was used to collect and analyze the temporal gait cycle characteristics which included the stride and the stance times for each leg. *Treadmill:* The subjects walked on a motorized treadmill (312-C, Bodyguard, Canada). *EMG:* A commercial wireless EMG system (Delsys Inc., Boston, MA) was used to collect data. *Harness:* All subjects wore a chest harness connected to a body weight supporting system (LiteGait, Mobility Research, Tempe, Arizona) for safety.

Experimental Design: Subjects were randomly divided into either a non-VR group or a VR group. Each subject underwent four conditions of treadmill walking at a self-selected pace. Prior to testing, each subject was asked to walk on the motorized treadmill and was instructed to walk at a comfortable pace that could be easily maintained for a long time. When the subject informed the investigator that such a speed had been reached, the value was recorded and was used for all testing conditions as the self-selected pace (SSP). The subjects then walked on the treadmill at their SSP for a familiarization period of approximately six minutes, which was considered sufficient for the achievement of reliable measurements (Matsas et al., 2000). During this period data was not recorded. The familiarization period was followed by the baseline period in which subjects in the VR group walked for 5 minutes in a VR environment and subjects in the non-VR group walked for 5 minutes in a Non-VR environment (a static image of the virtual corridor). This was followed by the adaptation period in which all subjects walked with the ankle weight attached to one leg for five minutes in either a VR environment or a Non-VR environment. This was followed by the washout period in which all subjects walked on the treadmill for five minutes without the ankle weight (VR group subjects were exposed to VR in this period). All subjects were instructed to keep their heads as steady as possible and maintain a steady focus at the center of the screen. All subjects were given the option to rest between conditions. During the experiment the lights in the room were switched off so that the subjects could concentrate only on the screen in front of them.

Data Recording and Processing: Data was recorded during each of the five minutes of baseline, loading and washout periods during treadmill walking. Kinematic Data: Before starting to walk on the treadmill, the foot-switch insoles of the Stride Analyzer were inserted into the subject's shoes after matching the size of the insoles to the subject's shoe size. The cables from the insoles were connected to a data logger that the subject carried around the waist (figure 1). Data was recorded through the data logger onto a memory card. EMG electrodes were attached to the muscle bellies of the following four muscles on each lower limb: tibialis anterior (LTA and RTA), medial head of gastrocnemius (LGASTRO and RGASTRO), vastus medialis (LVM and RVM) and biceps femoris (LHAMS and RHAMS). Cables connecting these electrodes were attached to an EMG data logger that was also attached to the waist of subject using a Velcro waistband (figure 1). Data was wirelessly sent from the data logger to a laptop for collecting EMG data. An external trigger was used to synchronize the data recorded from the stride analyzer and the wireless EMG systems.

The stride analyzer software was used to process the data from the stride analyzer. The data obtained included swing and stance times of the right and the left limbs. Cadence (the number of steps per minute) was calculated by adding the number of stances of both the sides and dividing by five (each of the three periods – baseline, adaptation and washout, were five minutes long). The loading effect on cadence was calculated by subtracting the cadence during baseline from the cadence during the adaptation period, while the unloading effect was determined by subtracting the cadence during adaptation from that during the washout period. The cadence measure was an average measure for the entire period. Regarding swing and stance times, after deleting the first and last five swing and stance times (because of acceleration and deceleration effects) in each of the three periods, the first ten (early period) and the last ten (late period) swing and stance times, were used for data analysis. Two adaptive effects were tested for the swing and stance times – short term and long term. Short-term adaptation comprised of the *loading effect* (defined as the difference between the early adaptation and the late baseline periods) and the *unloading effect* (defined as the difference between the early washout and the late adaptation periods). The *long-term adaptation effect* is defined as the difference between the late adaptation and the early adaptation periods. EMG data from the four muscles were recorded using the EMG software (Delsys data acquisition software). Data was collected at 1000 Hz and normalized to maximum voluntary contraction for each muscle for each subjects. Prior to normalization the data was rectified, filtered, using a butterworth bandpass filter (20-450Hz) and moving window averaged (50ms window width). In each of the three periods, early and late EMG effects were defined as the EMG activity between the first ten (early effects) and the last ten (late) steps, after removing the first five and the last five gait cycles (to remove the effects of acceleration and deceleration). Within these periods, the EMG data was further integrated. The EMG *loading effect* was defined as the difference between the normalized integrated EMG (IEMG) for the early adaptation period and the late baseline period, The EMG *unloading effect* was defined as the difference between the normalized integrated EMG (IEMG) for the early washout and the late adaptation periods, and the EMG *long-term adaptation effect* was defined as the difference between the late adaptation and the early adaptation periods.

Statistical Analysis: *Kinematic Variables:* To determine changes in cadence across the groups, a Mixed factor ANOVA was performed with the repeating factor being locomotor period (baseline, adaptation and washout) and group (VR and Non-VR) being the between subjects factor. To determine the *loading effect* (adaptation period – baseline period) and the *unloading effect* (washout period – adaptation period) of VR on cadence for unilateral limb loading, separate univariate ANOVAs were performed.

Regarding the two kinematic temporal variables (Swing time and Stance time), a mixed factor ANOVA was used to determine significant overall effect of the repeating factor (Locomotor period: Baseline, Adaptation Early, Adaptation Late, Washout Early and Washout Late) across group (VR and Non-VR), kinematic variables (Swing time and Stance time) and laterality (Right and Left). To determine the short term effect of VR on limb loading and unloading and the long-term adaptive effect of VR on limb loading, unpaired t-tests were performed for swing and stride times across left and right sides on the absolute differences between early adaptation and late baseline periods *(loading effect)*, early washout and late adaptation periods *(unloading effect)* and late adaptation and early adaptation periods *(long-term adaptation effect*). The level of significance was set at $\alpha = 0.05$.

EMG Variables: To determine the EMG correlates of the effect of VR on locomotor adaptation, mixed factor ANOVA was performed with locomotor period as the repeating factor (5 levels: Baseline, Adaptation Early, Adaptation Late, Washout Early and Washout Late) across group (VR and Non-VR) and laterality (Right and Left) for the following four muscles (tibialis anterior - TA, medial head of gastrocnemius - GASTRO, vastus medialis – VM and biceps femoris - HAMS). To determine the short term effect of VR on limb loading and the adaptive effect of VR on limb loading, unpaired t-tests were performed for each of the four muscles across left and right sides on the absolute differences between early adaptation and baseline periods (*loading effect*), early washout and late adaptation periods *(unloading effect)* and late adaptation and early adaptation periods *(long-term adaptation effect*). The level of significance was set at $\alpha = 0.05$. All the statistical analyses were performed using the SPSS (version 18.0, Chicago, IL, USA) statistical package.

RESULTS

The objective of this study was to determine whether a VR environment providing self perception of motion during treadmill walking influences the motor adaptation of a locomotor task (unilateral limb loading) using kinematic variables and EMG correlates. Although the SSP of the VR group was slightly higher (1.01±0.29 meters/second) than the SSP of the Non-VR group (0.94±0.32), there was no significant difference between the two groups (p=0.34).

Kinematic Variables: Data from two of the subjects was not recorded in the stride analyzer system due to technical errors therefore the final analysis of kinematic variables were from seven subjects in each of the Non-VR and VR groups. Regarding cadence, the mixed factor ANOVA revealed a significant interaction between group and locomotor periods ($F_{2,22}$=2.721, p=0.04). This means that the change in cadence caused by unilateral limb loading was significantly different between the VR and the Non-VR groups. There was a significant *loading effect* (adaptation – baseline periods) of VR over the Non-VR group ($F_{1,11}$=3.912, p=0.037). In order to adapt to the load, subjects exposed to VR reduced the number of steps by 10.76% whereas subjects not exposed to VR, increased the number of steps by 2.37% (figure 2). The *unloading effect* (washout – adaptation periods) of VR over the Non-VR groups did not reach significance ($F_{1,11}$=2.16, p=0.085).

Regarding swing and stance times, the mixed factor ANOVA revealed an overall effect of locomotor period ($F_{4, 28}$=2.888, p=0.012). This effect was significant across group, i.e., there was a significant interaction between locomotor period and group ($F_{4, 28}$=2.204, p=0.035) which means that adapting/de-adapting to the load was significantly different for the Non-VR and the VR groups. There was no significant interaction between the locomotor period and limb laterality ($F_{4, 28}$=0.04, p=0.498) but the interaction between locomotor adaptation and the two kinematic temporal variables was significant ($F_{4,28}$=2.326, p=0.029). To determine the short-term *loading effect*, unpaired t-tests were performed for each of the two kinematic temporal variables (Swing time and Stance time) across left and right sides on the absolute differences between early loading and baseline periods. No significant differences were demonstrated for the early effects of loading on the temporal kinematic variables. To determine the short-term *unloading effect*, unpaired t-tests were performed for each of the two kinematic temporal variables (Swing time and Stance time) across left and right sides on the absolute differences between early washout and late loading periods (figure 3). A significantly larger decrease occurred for the left swing phase (p=0.035) and the right stance phase (p=0.045), which was 9.18% and 4.38% respectively for the VR group and 3.26% and 2.36% for the Non-VR group. A significantly smaller increase occurred for the right swing phase (p=0.049) for the VR group (1.73%) in comparison to the Non-VR group (6.32%). The *unloading effect* for the left stance phase was not significant (p=0.34).

(INSERT FIGURE 2 HERE)

(INSERT FIGURE 3 HERE)

To determine the *long-term adaptation effect*, unpaired t-tests were performed for each of the two kinematic temporal variables (Swing time and Stance time) across left and right sides on the absolute differences between late adaptation and early adaptation periods. A significantly larger increase occurred for the right swing phase (p=0.029) which was 3.15% for the VR group as compared to 1.31% for the Non-VR group. The *long-term adaptation effect* for the right stance phase was approaching significance (p=0.071) with a smaller change in the VR group (2.64%) in comparison to the Non-VR group (6.38%). The *long-term adaptation effects* for the left swing and stance times were not significant (p=0.328 and 0.300, respectively).

EMG Variables: The mixed factor ANOVA revealed an overall effect of locomotor period ($F_{4,28}$=10.298, p=0.000) and this effect was significant across group, i.e., there was a significant interaction between locomotor period and group ($F_{4,28}$=7.412, p=0.000), which means that adapting to the load was significantly different for the Non-VR and the VR groups. The adaptation was not significantly different across limb laterality ($F_{4,28}$=0.263, p=0.901) or different muscle types ($F_{12,84}$=1.066, p=0.387). To determine the *short term loading effect* unpaired t-tests were performed for each of the four muscles across left and right sides on the absolute differences between early loading and baseline periods. Although a trend for higher levels of IEMG activity were noted for all muscles in the Non-VR group in comparison to the VR group (figure 4), a significant difference was demonstrated only for RVM (p=0.041). The RVM activity was significantly higher for the Non-VR group (89.21%) than the VR group (42.56%). Two muscles, RTA (p=0.100) and LHAMS (p=0.084) were shown to approach significance. To determine the *short term unloading effect*, unpaired t-tests were performed for each of the four muscles across left and right sides on the absolute differences between early washout and late loading periods. Only the RHAMS (p=0.042) showed a significantly higher activity for the Non-VR group (30.5%) than the VR group (24.46%, figure 5).

(INSERT FIGURE 4 HERE)

(INSERT FIGURE 5 HERE)

To determine the *long-term adaptation effect*, unpaired t-tests were performed for each of the four muscles across left and right sides on the absolute differences between late adaptation and early adaptation periods. Only the RGASTRO (p=0.032) showed a significantly higher activity for the Non-VR group (39.67%) than the VR group (24.47%). Another muscle, RTA (p=0.081) was shown to approach significance.

DISCUSSION

The objectives of the study were to determine whether a VR environment providing self perception of motion during treadmill walking with a unilateral limb weight would produce short-term *loading* and *unloading* effects and *long-term adaptation* effects in temporal gait patterns and muscle activity different from the Non-VR condition. Significantly different effects of VR on locomotor adaptation were demonstrated in terms of cadence, swing time, stance time and EMG activity in comparison to the Non-VR group.

The effect of VR on cadence. Shorter stride lengths have been shown to occur during limb loading with ankle weights (Claremont and Hall, 1988) in comparison to the unloaded condition. If the velocity of treadmill walking stays the same, this would mean more frequent steps than the unloaded condition. This was our finding for the *loading effect* on cadence in the Non-VR group. Therefore in the absence of optic flow, the effect of limb loading was to increase cadence. However, in the presence of self perception of motion (VR group), cadence reduced during limb loading. This difference in effect could be explained by relating cadence to stride length given the velocity of walking is the same. In stressful situations walking becomes stiffer – shorter and more frequent steps. Examples of such situations are walking with a load (Claremont and Hall, 1998, Miller and Stamford, 1987; Skinner and Barrack, 1990), backward walking (Thortensson, 1986; Winter et al., 1989), fear of falling in older adults (Maki, 1997) and walking on slippery floors (Cham and Redfern, 2002). The task of lifting a load and walking on a treadmill may have made the subjects more cautious and the VR environment, by providing self perception of motion, may help to reduce the need to employ a more cautious gait.

The effect of VR on swing and stride times. Asymmetrical limb loading using ankle weights has been shown to affect swing and stride times (Skinner and Barrack, 1990). In that study it was shown that the loaded limb had increased swing phase and reduced stance phase and the unloaded limb had reduced swing phase and increased stance phase. When the load was removed there is a reversal of the effects i.e., the loaded limb had reduced swing phase and increased stance phase and the unloaded limb had increased swing phase and reduced stance phase (Smith and Martin, 2007). Even when both limbs are loaded, there is an increase in swing time and reduction in stance time (Eke-Okoro et al., 1985). This was also our finding for both the VR and the Non-VR groups. However we were interested on the effect of VR on the magnitudes of these changes. When the magnitudes of these changes (figure 3) were compared, significant differences between the two groups were demonstrated. In the study by Smith and Martin (2007), it was shown that healthy human subjects adapted quickly to unilateral limb loading and unloading. Therefore, in contrast to unloading effect where we found several differences, the *long-term adaptation effect* was shown only in one variable. The reason why several significant differences were observed for the *unloading effect* but not for the *loading effect* could be because the magnitude of variability (standard deviation) was higher for the *loading effect*. In terms of the magnitudes of these changes, we found that the perception of self motion enhanced these normal responses to limb loading for the loaded swing and the unloaded stance phases and reduced the effect on the unloaded swing phase.

The effect of VR on EMG activity. Past studies have shown increased energy expenditure as a result of weighted walking (Miller and Stamford, 1987; Skinner and Barrack, 1990). Skinner and Barrack (1990) showed that asymmetrical loading with ankle weights of 4 lbs caused an increased energy expenditure of 7% over unloaded walking. Considering that the weight used in this study was 10lbs, the energy expenditure would be higher with increased levels of muscle activity. The relationship between muscle activity and loading during treadmill walking has also been shown in the past (Bachmann et al., 2008). In that study, it was shown that loading the body caused an increase in EMG activity while unloading the body using a bodyweight support system reduced the muscle activity. Could visual feedback of self motion influence this muscle activity during loading? Indeed, overall muscle activity was significantly reduced by VR. In terms of individual muscle activity, muscles of the thigh region - the RVM (*loading effect*) and the RHAMS (*unloading effect*) and RGASTRO (*long-term adaptation effect*) showed significantly more activity in the Non-VR group. These changes were observed only on the right side which may be the reason for larger differences on the unloaded right side (both swing and stance times were different) as opposed to the loaded left side (only swing time was different). Ankle loading has been shown to modulate both hip kinetics and EMG during locomotion (Gordon et al., 2009). And these changes have been observed both in the ipsilateral and contralateral sides. However, little or no EMG changes were observed on the contralateral unloaded side of spinal cord injured subjects (Ferris et al., 2004, Gordon et al., 2009). This demonstrates that inter-limb coordination during locomotion with unilateral limb loading may be under supraspinal control and therefore, sensory feedback (e.g. optic flow) could have influenced such a coordination task.

Why should muscle activity be higher when the perception of self motion is reduced or lacking? The reduction of self perception of motion in the Non-VR group may have caused the higher brain centers to overcompensate for the load during treadmill walking demonstrated through increased muscle activity. Why should this be so? Since the same task can be achieved with less muscle activity, longer strides, and less number of steps (VR group), it is probably the perception of self motion that provides a better scale of the locomotor requirements. When a subject is on the treadmill, the subject needs to gauge the velocity of the treadmill correctly to walk normally. This was easily achieved by all the subjects whether in the VR or the Non-VR group. Once the load was attached, a second task needed to be solved – determining the weight of the load so that the loaded limb can be carried at the previous SSP of the treadmill. Understanding the dynamics of the new system was a problem that needed extra cues to solve. In the Non-VR group, the subject relied on the kinesthetic receptors to provide accurate feedback to solve both problems – not an easy task. Although they had vision during task performance, the ability to perceive self perception of motion was diminished as they looked straight ahead at the screen with a static image of the corridor. This may have caused the difficulty in correctly estimating the number of steps required - the subjects therefore, walked with more steps (increase in cadence) and since the velocity was the same, with shorter stride lengths. More steps in the same time would mean more frequent joint movements and more overall muscle activity (demonstrated by an increasing direction in muscle activity). In contrast, subjects in the VR group had one less task to solve – determining the velocity of the treadmill was easily achieved using visual feedback by the VR environment which provided the perception of self motion. It was well known that in situations of sensory conflict, the brain relies more on the visual system (Bagesteiro et al., 2006; Flanagan and Rao, 1995; Wolpert et al., 1994; Wolpert et al., 1995). Therefore the determination of the load dynamics during walking was the only task that the kinesthetic receptors had to provide feedback for in the VR group. The resolution of such novel sensorimotor situations may be easier if different sensory modalities have different feedback roles.

One of the limitations of this study was the use of the same weight for all subjects although they varied in height and weight. However the subjects had an average weight of 158.75lbs and an ankle weight of 10 lbs was around 6.3% of the body weight. Based on past literature, this was considered sufficient to perturb gait and cause locomotor adaptation (Byrne et al., 2002; Skinner and Barrack, 1990; Miller and Stamford, 1987).

In this study it was shown that optical flow from the VR environment influenced locomotor adaptation. It was assumed that the VR simulation caused perception of self motion which was the cause of the observed effects. Whether such an effect can also be caused by the VR environment moving in the opposite direction or in an orthogonal direction or by any other environment, remains to be investigated. This will be the course of future studies. Another area of future investigation would be to study the VR effect when it is yoked to the subject's motion to make it a true feedback signal. It should also be noted that the ability to perceive self-motion is affected by the optic flow of the environment (Durgin et al., 2007) and in addition the preferred walking speed is also influenced by optic flow (Mohler et al., 2007). However we expected that these phenomena were controlled when we kept the treadmill velocity constant at the subjects' preferred walking speed and matched the optic flow to that speed.

In summary, healthy human subjects walked on a treadmill with a load attached on one leg either without or in a virtual condition which provided perception of self motion. Kinematic and EMG data demonstrated significant differences in specific variables which indicated that locomotor adaptation may benefit from visual cues providing perception of self motion. Specifically, the perception of self motion reduced cadence, reduced overall muscle activity, while enhancing the normal responses to limb loading for the loaded swing and the unloaded stance phases and reducing the effect on the unloaded swing phase. The lack of visual cues may lead the subjects to adopt specific strategies (higher muscle activity, shorter and more frequent steps, less increase in loaded swing time and unloaded stance time) when faced with situations which require adaptive strategies. VR technology, providing such perceptual cues, has an important role to play in enhancing locomotor adaptation. These findings may have a significant impact on neuro-rehabilitation programs. Further studies will explore the mechanism behind such adaptations and also investigate such adaptive correlates in the elderly and patient populations.

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Figures

Figure 1A. One subject walking on a treadmill wearing the safety harness while being exposed to the virtual corridor. The figure also shows the data loggers of the wireless EMG system and the Stride Analyzer strapped to the back of the subject. **B.** A close-up view of the ankle weights worn on one leg during the locomotor adaptation experiments.

Figure 2. The change in cadence as a result of VR exposure on unilateral limb loading during treadmill walking (in steps/min) shown as the difference between the adaptation and the baseline periods (*loading effect*) and the washout and the adaptation periods (*unloading effect*) for the two VR groups. Error bars are standard deviation. *$p<0.05$. There was a significant *loading effect* (adaptation – baseline periods) of VR over the Non-VR group. In order to adapt to the load, subjects exposed to VR reduced the number of steps by 10.76% whereas subjects not exposed to VR, increased the number of steps by 2.37%. The *unloading effect* (washout – adaptation periods) of VR over the Non-VR groups did not reach significance.

Figure 3. The magnitude of the short-term unloading effect on Swing and Stride times (in seconds) shown as differences between the early washout period and the late adaptation period for both the limbs for the two VR groups. Error bars are standard deviation. *$p<0.05$. A significantly larger *unloading effect* occurred for the left swing phase and the right stance phase, which was 9.18% and 4.38% respectively for the VR group and 3.26% and 2.36% for the Non-VR group. A significantly smaller *unloading effect* occurred for the right swing phase for the VR group (1.73%) in comparison to the Non-VR group (6.32%). The *unloading effect* for the left stance phase was not significant.

Figure 4. The change in normalized integrated EMG activity as a result of VR exposure on unilateral limb loading during treadmill walking shown as the difference between the early

loading and the baseline periods (*loading effect*) for the two VR groups. RTA – right tibialis anterior, RGASTRO – right medial head of gastrocnemius, RVM – right vastus medialis, RHAMS – right hamstrings (biceps femoris), LTA – left tibialis anterior, LGASTRO – left medial head of gastrocnemius, LVM – left vastus medialis and LHAMS – left hamstrings (biceps femoris). Error bars are standard deviation. Although a trend for higher levels of activity were noted for all muscles in the Non-VR group in comparison to the VR group, a significant difference was demonstrated only for RVM. The RVM activity was significantly higher for the Non-VR group (89.21%) than the VR group (42.56%). Two muscles, the RTA and LHAMS were shown to approach significance.

Figure 5. The change in normalized integrated EMG activity as a result of VR exposure on unilateral limb loading during treadmill walking, shown as the difference between the early washout and the late loading periods (*unloading effect*) for the two VR groups. Error bars are standard deviation. Only the RHAMS showed a significantly higher activity for the Non-VR group (30.5%) than the VR group (24.46%).