The Influence of Visual Perception of Self-Motion on Locomotor Adaptation to 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 the present study, the authors 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 3 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 that require adaptations. VR technology, providing such perceptual cues, has an important role in enhancing locomotor adaptation.

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

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 or task constraints (Gordon, Fletcher, Melvill Jones, & Block, 1995; Lam, Anderschitz, & Dietz, 2006; Morton & Bastian, 2006; Reisman, Wityk, Silver, & Bastian, 2007; Richards, Mulavara, & Bloomberg, 2007). 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, virtual information is given precedence (Bagesteiro, Sarlegna, & Sainburg, 2006; Flanagan & Rao, 1995; Wolpert, Ghahramani, & Jordan, 1994, 1995). Visual sensory information may therefore play a key role in learning or relearning locomotor dynamics.

In daily life, humans 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 is improved, which leads to an improvement in the quality of life. Age- and disease-related asymmetrical gait patterns (Hsu, Tang, & Jan, 2003; Lamontagne & Fung, 2004; Olney, Griffin, & McBride, 1994; Plotnik, Giladi, & Hausdorff, 2007) 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 (Choi & Bastian, 2007; Reisman, Wityk, Silver, & Bastian, 2007), (b) a robotic device that is attached to the lower limbs and can provide resistance during treadmill walking (Lam, Anderschitz, & Dietz, 2006), (c) actuators that provide assistance or resistance to the swing leg during treadmill training (Kurz & Stergiou, 2007), and (d) by unilateral limb loading with ankle weights that provide resistance to the swinging leg (Byrne, Stergiou, Blanke, Houser, Kurtz, & Hageman, 2002).

Unilateral limb loading is a type of adaptive activity that causes changes in spatiotemporal gait patterns (Byrne et al., 2002; Skinner & Barrack, 1990) as well as physiological changes (Claremont & Hall, 1988; Miller & 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 & 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 (Bachmann, Müller, van Hedel, & Dietz, 2008; Stephens & Yang, 1999). 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, Wu, Kahn, Daher, & Schmit, 2009). It has been suggested that the CNS has the ability to control timing and amplitude of muscular activity during limb loading (Stephens & Yang, 1999); however, it is not clear whether external sensory feedback has the ability to influence this control.

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Unilateral limb loading has been used as a therapeutic tool to reduce abnormal limb movement and improve stability (Hewer, Cooper, & Morgan, 1972; Morgan, 1975). It has been used in the past to simulate paretic gait in normal subjects (Eke-Okoro, Larsson, & Sandlund, 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, Lee, Charalambous, & Vrongistinos, 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, such as those with stroke. However, this task may be made easier if optic flow (OF) were provided to the subjects. It is known that OF providing perception of self-motion affects several spatiotemporal patterns during treadmill walking (Katsavelis, Mukherjee, Decker, & Stergiou, 2010a, 2010b; Prokop, Schubert, & Berger, 1997; Varraine, Bonnard, & Palhous, 2002). 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 that usually occur in static environments like walking on a treadmill in a room, providing perception of self-motion is not an easy task. Virtual reality (VR) technology that provides perception of self-motion may be the answer to this problem.

VR 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, Pick, Ashmead, & Garing, 1995). Simply manipulating the visual scene’s velocity was shown to affect the preferred walking speed (Mohler, Thompson, Creen-Regehr, Pick, & Warren, 2007) and the segmental kinematics during steady standing (Dokka, Kenyon, & Keshner, 2009). In the present research study we 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 VR environment. All subjects adapted to unilateral loading with an ankle weight while walking on a treadmill at a self-selected pace. The 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 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 (intuitive 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 use 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 overground walking with ankle loading.

Method

Subjects

Sixteen healthy adults (7 men and 9 women) participated in the study (M age = 28.13 ± 3.67 years; M height = 171.25 ± 9.50 cm; M weight = 158.75 ± 22.17 lbs). 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 99 × 68 inch flat screen (Stewart Filmscreen Corp., Torrance, CA) that was positioned 1.5 m away from the plane of motion. This created a vertical and horizontal field of view of 59.39° and 79.42°, 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 m 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 subsequently in the Experimental Design subsection), 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 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 (Jäncke, Cheetham, & Baungartner, 2009), which
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**FIGURE 1.** (A) 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.

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, Hartmann, Böcking, Vorderer, Klimmt, Schramm, Sari 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, Mukherjee, Decker, & Stergiou, 2010a, 2010b). The presence questionnaire (Witmer & 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.

**Weight**

An ankle weight equivalent to 10 pounds 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, Stergiou, Blanke, Houser, Kurz, & Hagemann, 2002; Miller & Stamford, 1987; Skinner & Barrack, 1990).

**Stride Analyzer**

A commercial footswitch system–stride analyzer (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).

**Electromyography**

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, AZ) 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 (SSP).
Before 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 SSP. The subjects then walked on the treadmill at their SSP for a familiarization period of approximately 6 min, which was considered sufficient for the achievement of reliable measurements (Matsas, Taylor, McBurney, & Knee, 2000). During this period, data were not recorded. The familiarization period was followed by the baseline period in which subjects in the VR group walked for 5 min in a VR environment and subjects in the non-VR group walked for 5 min 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 5 min 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 5 min 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 were recorded during each of the 5 min of baseline, loading and washout periods during treadmill walking. 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 were 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 (TA; LTA and RTA), medial head of gastrocnemius (GASTRO; LGASTRO and RGASTRO), vastus medialis (VM; LVM and RVM), and biceps femoris (HAMS; LHAMS and RHAMS). Cables connecting these electrodes were attached to an EMG data logger that was also attached to the waist of the subject using a Velcro waistband (Figure 1). Data were 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 5 (each of the three periods—baseline, adaptation, and washout—were 5 min 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 (early period) and last 10 (late period) swing and stance times were used for data analysis. Two adaptive effects were tested for the swing and stance times—short and long term. Short-term adaptation comprised the loading effect (the difference between the early adaptation and the late baseline periods) and the unloading effect (the difference between the early washout and late adaptation periods). The long-term adaptation effect was defined as the difference between the late adaptation and the early adaptation periods. EMG data from the four muscles were recorded using the EMGWorks (Delsys, Inc., Boston, MA). Data were collected at 1,000 Hz and normalized to maximum voluntary contraction for each muscle for each subject. Before normalization the data were rectified, filtered, using a butterworth bandpass filter (20–450 Hz) and moving window averaged (50 ms window width). In each of the three periods, early and late EMG effects were defined as the EMG activity between the first (early) and last 10 (late) steps, after removing the first and last five gait cycles (to remove the effects of acceleration and deceleration). Within these periods, the EMG data were 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 analysis of variance (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–baseline period) and the unloading effect (washout–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),...
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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 and early adaptation periods (long-term adaptation effect). The level of significance was set to \( \alpha = .05 \).

**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 (baseline, adaptation early, adaptation late, washout early, washout late) with locomotor period as the repeating factor (baseline, adaptation early, adaptation late, washout early, washout late) and laterality (right and left). 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 the left and right sides on the absolute differences between the early washout 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 the left and right sides on the absolute differences between early washout and late adaptation periods (Figure 3). A significantly larger decrease occurred for the left swing phase \( (p = .035) \) than the right stance phase \( (p = .045) \), which was 9.18% and 4.38% for the VR group and 3.26% and 2.36% for the non-VR group, respectively. A significantly smaller increase occurred for the right swing phase \( (p = .049) \) than the left stance phase \( (p = .034) \), which was 6.32% in comparison with the non-VR group (6.32%). The unloading effect for the left stance phase was not significant \( (p = .34) \).

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 the left and right sides on the absolute differences between late and early adaptation periods. A significantly larger increase occurred for the right swing phase \( (p = .029) \), which was 3.15% for the VR group compared with 1.73% for the non-VR group and 1.31% for the VR group compared with 0.94% for the non-VR group. The long-term adaptation effect for the right stance phase was approaching significance \( (p = .071) \) with a smaller change in the VR group (2.64%) in comparison to the non-VR group (6.8%). The long-term adaptation effect for the left swing and stance times were not significant \( (p = .328 \text{ and } .300, \text{ respectively}) \).

**Kinematic Variables**

Data from two of the subjects were 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 period \( (F(2, 22) = 2.721, \ p = .04) \). 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 period) of VR over the non-VR group \( (F(1, 11) = 3.912, \ p = .037) \). 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 \( (F(1, 11) = 2.16, \ p = .085) \).

Regarding swing and stance times, the mixed-factor ANOVA revealed an overall effect of locomotor period \( (F(4, 28) = 2.888, \ p = .012) \) for the following four muscles (TA, GASTRO, VM, HAMS). The mixed-factor ANOVA revealed an overall effect of VR over the non-VR group \( (F(4, 28) = 2.204, \ p = .035) \), which means that adapting and deadapting to the load was significantly different for the non-VR and VR groups. There was no significant interaction between the locomotor period and limb laterality, \( (F(4, 28) = 0.040, \ p = .498) \) but the interaction between locomotor adaptation and the two kinematic temporal variables was significant \( (F(4, 28) = 2.326, \ p = .029) \). 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 the left and right sides on the absolute differences between the early washout and late adaptation periods (Figure 3). A significantly larger decrease occurred for the left swing phase \( (p = .035) \) and the right stance phase \( (p = .045) \), which was 9.18% and 4.38% for the VR group and 3.26% and 2.36% for the non-VR group, respectively. A significantly smaller increase occurred for the right swing phase \( (p = .049) \) for the VR group (1.73%) in comparison with the non-VR group (6.32%). The unloading effect for the left stance phase was not significant \( (p = .34) \).

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 the left and right sides on the absolute differences between late and early adaptation periods. A significantly larger increase occurred for the right swing phase \( (p = .029) \), which was 3.15% for the VR group compared with 1.73% for the non-VR group and 1.31% for the VR group compared with 0.94% for the non-VR group. The long-term adaptation effect for the right stance phase was approaching significance \( (p = .071) \) with a smaller change in the VR group (2.64%) in comparison to the non-VR group (6.8%). The long-term adaptation effect for the left swing and stance times were not significant \( (p = .328 \text{ and } .300, \text{ respectively}) \).

**EMG Variables**

The mixed-factor ANOVA revealed an overall effect of locomotor period \( (F(4, 28) = 10.298, \ p = .000) \) and this effect was significant across group \( (i.e., \ there \ was \ a \ significant \ interaction \ between \ locomotor \ period \ and \ group; \ F(4, 28) = 7.412, \ p = .000) \), 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 \( (F(4, 28) = 0.263, \ p = .901) \) or different
FIGURE 2. The change in cadence as a result of virtual reality (VR) exposure on unilateral limb loading during treadmill walking (in steps/minute) 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 < .05). There was a significant loading effect of VR over the non-VR group. 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 of VR over the non-VR groups did not reach significance.

FIGURE 3. The magnitude of the short-term unloading effect on swing and stance times (in seconds) shown as differences between the early washout period and the late adaptation period for both the limbs for the two virtual reality (VR) groups. Error bars are standard deviation (*p < .05). A significantly larger unloading effect occurred for the left (L) swing and the right (R) stance phases, 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 R swing phase for the VR group (1.73%) in comparison with the non-VR group (6.32%). The unloading effect for the left stance phase was not significant.
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muscle types \( F(12, 84) = 1.066, p = .387 \). To determine the short-term loading effect, unpaired \( t \)-tests were performed for each of the four muscles across the 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 with the VR group (Figure 4), a significant difference was demonstrated only for RVM \( (p = .041) \). The RVM activity was significantly higher for the non-VR group (89.21\%) than the VR group (42.56\%). Two muscles, RTA \( (p = .100) \) and LHAMS \( (p = .084) \) were shown to approach significance. To determine the short-term unloading effect, unpaired \( t \)-tests were performed for each of the four muscles across the left and right sides on the absolute differences between the early washout and late loading periods. Only the RHAMS \( (p = .042) \) showed a significantly higher activity for the non-VR group (30.5\%) than the VR group (24.46\%; Figure 5).

To determine the long-term adaptation effect, unpaired \( t \)-tests were performed for each of the four muscles across the left and right sides on the absolute differences between late adaptation and early adaptation periods. Only the RGASTRO \( (p = .032) \) showed a significantly higher activity for the non-VR group (39.67\%) than the VR group (24.47\%). Another muscle, RTA \( (p = .081) \) 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 & 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

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**FIGURE 4.** The change in normalized integrated electromyography (IEMG) activity as a result of virtual reality (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. 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 with the VR group, a significant difference was demonstrated only for right vastus medialis (RVM). The RVM activity was significantly higher for the non-VR group (89.21%) than for the VR group (42.56%). Two muscles, the right tibialis anterior (RTA), and left hamstring (biceps femoris; LHAMS) were shown to approach significance. RGASTRO = right medial head of gastrocnemius; RHAMS = right hamstring (biceps femoris); LTA = left tibialis anterior; LGASTRO = left medial head of gastrocnemius; LVM = left vastus medialis.
stiffer—shorter and more frequent steps. Examples of such situations are walking with a load (Claremont & Hall, 1998; Miller & Stamford, 1987; Skinner & Barrack, 1990), backward walking (Thortensson, 1986; Winter, Pluck, & Yang, 1989), fear of falling in older adults (Maki, 1997), and walking on slippery floors (Cham & 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 & 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 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; Smith & Martin, 2007). Even when both limbs are loaded, there is an increase in swing time and reduction in stance time (Eke-Okoro, Larson, & Sandlund, 1985). This was also our finding for both the VR and the non-VR groups. However, we were interested in 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 the 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.

Effect of VR on EMG Activity

Past studies have shown increased energy expenditure as a result of weighted walking (Miller & Stamford, 1987; Skinner & Barrack, 1990). Skinner and Barrack 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 10 pounds, 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, Müller, van Hedel, & Dietz, 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 (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 (K. E. Gordon, Wu, Kahn, Dhaher, & Schmit, 2009). 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, Gordon, Beres-Jones, & Harkema, 2004; Gordon, Wu, Kahn, Dhaher, & Schmit, 2009). This demonstrates that interlimb coordination during locomotion with unilateral limb loading may be under supraspinal control and therefore, sensory feedback (e.g., OF) 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? Because the same task can be achieved with less muscle activity, longer strides, and fewer 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 of the subjects, whether in the VR or the non-VR groups. Once the load was attached, a second task needed to be solved—determining the weight of the load so that the loaded limb could 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 might have caused the difficulty in correctly estimating the number of steps required—therefore, the subjects walked with more steps (increase in cadence) and, because 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 is well known that in situations of sensory conflict, the brain relies more on the visual system (Bagestead, Sarlegna, & Sainburg, 2006; Flanagan & Rao, 1995; Wolpert, Ghaehramani, & Jordan, 1994, 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 158.75 pounds and an ankle weight of 10 pounds was around 6.3% of the body weight. Based on past literature, this was considered sufficient to perturb gait and cause locomotor adaptation (Byrne, Stergiou, Blanke, Houser, Kurz, & Hagemann, 2002; Miller & Stamford, 1987; Skinner & Barrack, 1990).

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 further 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, Reed, & Tigue, 2007) and in addition, the preferred walking speed is also influenced by optic flow (Mohler, Thompson, Creem-Regehr, Pick, & Warren, 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 that 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 and 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 that 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 neurorehabilitation programs. Future investigations should explore the mechanism behind such adaptations.
and also investigate such adaptive correlates in the elderly and patient populations.

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**NOTE**

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

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