Movement Strategies for Maintaining Standing Balance during Arm Tracking in People with Multiple Sclerosis

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The purpose of this study was to quantify hip and ankle movement strategies during a standing, arm tracking task in people with multiple sclerosis (MS). Full body kinematics and kinetics were assessed using motion analysis cameras and force plates in 9 MS and 9 age-matched controls. While standing, participants used their dominant hand to track a target moving around a large horizontal or vertical figure-eight on a screen in front of them. The target moved at constant speed, or linearly increasing speeds, with a frequency between 0.05 Hz and 0.35 Hz. Hip and ankle moments and angles during tracking were calculated from kinematic and kinetic measurements. The ratio of peak-to-peak (PP) hip/ankle moments (kinetics) and angles (kinematics) were calculated to determine the strategies of the hips and ankles used to maintain balance during arm movements. The center-of-mass (CoM) root mean square (RMS) acceleration was calculated as a measure of overall balance performance. The MS group produced larger PP hip/ankle moments at all speeds, compared with the control group (p<0.05). The CoM RMS acceleration increased with tracking speed for both groups but was not significantly different between groups. Additionally, the ratios of hip/ankle moments were highly correlated with the Berg Balance Scale during horizontal steady speed tracking in MS. These results suggest that people with MS increase the use of the hip during standing arm tracking compared to age-matched controls. This adapted strategy might allow people with MS to achieve similar balance performance to controls, possibly increasing the importance of the hip in maintaining balance during voluntary movements.
Introduction

An improved understanding of the movement strategies for standing balance in people with multiple sclerosis (MS) is potentially important for improving balance control and preventing falls. Balance impairments affect up to 82% of the MS population (Martyn and Gale, 1997), resulting in an increase in the number of falls (Cattaneo et al., 2002; Soyuer et al., 2006; Nilsagård et al., 2009; Matsuda et al., 2011; Prosperini et al., 2011; Sosnoff et al., 2011; Coote et al., 2012), even in the early stages of MS (Moen et al., 2011). The roles of the ankle and hip in balance control are interesting because two distinct strategies for maintaining balance, namely the ankle strategy and the hip strategy, have been proposed (Nashner and McCollum, 1985). These strategies might be used separately or together in varying degrees to produce optimal and adaptable balance control, depending on the difficulty of the balance task (Gatev et al., 1999; Runge et al., 1999; Hwang et al., 2009). Given the high incidence of falls, a better understanding of the ankle and hip movement strategies during standing balance in people with MS would provide insight into balance control problems and could aid in the planning of therapies to improve balance control.

Deficits in balance control have been speculated to stem from impaired sensory feedback and integration (Dietz, 1992; Horak et al., 1997) or related motor impairments associated with MS, such as spasticity (Rougier 2007). To date, sensorimotor impairments including impaired reflexes, spasticity, co-contraction (Toft et al., 1993; Crone et al., 1994; Sinkjaer et al., 1999; Morita et al., 2001), muscle fatigue, and torque asymmetries (Kalron et al., 2011), have been documented largely at the ankle. People with MS often have problems with ankle spasticity (Rizzo et al., 2004) that likely complicates balance control, since the ability to modulate ankle stiffness is an important component of the control of standing balance (Gatev 1999). Individuals
with MS show a correlation between level of spasticity, sway velocity, and amplitude of the soleus H-reflex (Sosnoff 2010). Hence, it is likely that impaired control at the ankle causes problems in the control of standing balance in people with MS, and balance performance must either be compromised, or a compensatory strategy, possibly involving hip control, is used to maintain balance.

Responding to an external balance perturbation requires a strategy for body movement that is likely affected by changes in neuromotor control associated with MS. Strategies that correct for balance perturbations are limited by external constraints imposed by the environment and task, and internal constraints imposed by the individual’s biomechanics (restricted joint range of motion, weakness) and nervous system (e.g. accuracy of sensory information, impairment to force and position control mechanisms) (Horak, 1996). In particular, MS causes delays in nerve signal conduction in sensory and motor pathways. Peripheral nerve conduction delays associated with diabetic neuropathy result in delayed postural responses (Inglis et al., 1994). Similarly, delays in sensory and motor conduction in spinal pathways (Pratt et al., 1992), or other delays in central processing (Shumway-Cook and Woollacott, 1985; Woollacott et al., 1986) can cause changes to balance strategies and movement corrections to external perturbations. In particular, sensory deficits such as a loss of proprioception or abnormal spatiotemporal coordination of automatic postural responses can cause reversals in the normal distal-to-proximal temporal sequencing of postural muscle activation, resulting in a decreased reliance on the ankle and an increased reliance on the hip for balance correction (Nashner et al., 1983; Di Fabio et al., 1990). In people with MS, it is likely that sensory and motor conduction delays result in changes to motor control strategies for balance correction.
Testing of balance control typically involves a perturbation (internal or external) with measurements of joint kinematics (range of motion) and kinetics (moment) to quantify the ensuing corrections. To identify the kinematic and kinetic patterns of the hip and ankle joints during standing balance, external perturbations to a standing platform are commonly applied (Horak et al., 1997). Several studies have addressed the effect of an external perturbation below or at the feet to characterize the role of ankle and hip joints in balance (e.g. Kuo and Zajac, 1993; Laessoe and Voigt, 2008; Terry et al., 2011). In contrast, the use of internal perturbations, such as arm movements, is somewhat less common, and control strategies include anticipatory balance corrections to account for the planned limb movement. Examining the effects of functional reach on balance control is important, as leg responses during reach not only stabilize the body but also initiate and assist whole-body reaching (Kaminski and Simpkins, 2001).

Often, people with MS are able to maintain balance in steady stance, but perform poorly when the balance task difficulty increases or if the task involves an internal perturbation such as functional reach (Van Emmerik et al., 2010). Thus, a benefit of identifying balance strategies during arm movement is that it models an important aspect of integrated functional movement.

In the current study, we recorded ankle and hip kinematic and kinetic data during standing, with an internal perturbation consisting of an arm tracking task. Data from participants with MS were compared with the performance of neurologically intact participants completing the same tasks. We hypothesized that the amount of hip relative to ankle movement during standing balance would increase to a greater extent in participants with MS compared with healthy controls.
Methods

Study Participants

Nine study participants with MS (age range: 38-57 yrs, mean age: 51 yrs) participated in this study. All participants with MS were community ambulators capable of walking 10 m and standing independently without assistive devices for up to one minute. At the time of this study, three of the nine participants were taking medication to improve their walking (dalfampridine: Ampyra; Acorda Therapeutics, Inc., Ardsley NY) and antispastic medication (baclofen: generic) to reduce the frequency and intensity of spasms. The clinical features of each participant are described in Table 1. Additionally, nine participants (age range: 43-57 yrs, mean age: 51.8 yrs) with no reported neurological damage were recruited for this study as controls. Exclusion criteria for this study included: significant cardiovascular problems, respiratory failure, major orthopedic problems including contracture of limbs, joint replacements, significant medical co-morbidity, concurrent illnesses limiting the capacity to conform to study requirements, or the inability to give informed consent. Informed consent was obtained prior to study participation and all procedures were conducted in accordance with the Helsinki Declaration of 1975 and approved by the Institutional Review Board of Marquette University and the Medical College of Wisconsin.

Clinical Assessment

Clinical tests were used to evaluate each MS participant’s function and ability to participate in the study. The following tests were conducted in participants with MS: the Modified Ashworth Scale (MAS) was used to measure muscle hypertonia (Bohannon and Smith,
1987), the Berg Balance Scale (BBS) was used to evaluate balance function (Berg et al., 1992)
the 25 feet walk test (25FWT) was used to evaluate walking ability (Kaufman et al., 2000), the
Fatigue Severity Scale (FSS) was administered to quantify general fatigue (Krupp et al., 1989),
the Symbol Digit Modalities Test (SDMT) was used to evaluate cognitive function
(Lewandowski, 1984) and the 9 hole peg test (9HPT) was used to evaluate motor coordination
(Goodkin et al., 1988). The 9HPT test was repeated twice for each hand, and the time required
for each iteration was recorded. The subjects were asked to self-identify their dominant hand.
All clinical tests were completed by a licensed physical therapist prior to beginning the
experiment. Participant-specific information is provided in Table 1.

**Experimental Setup**

Study participants were required to use their hand to track a moving target displayed and
controlled by LabVIEW (National Instruments Corp., Austin, TX) on a large screen in front of
them while standing. A SMARTboard (SMART Technologies, Calgary, Canada) (39 5/8” x 52
3/4”) was used to simultaneously display the tracking target and feedback of the finger position.
The position of the SMARTboard screen was adjusted such that when the participant’s arm was
fully extended (parallel to the ground) and pointing directly forward, the middle of the
SMARTboard screen would be centered and just out of reach of the subject’s index finger. The
size of the tracking trajectory was adjusted according to the length of the subject’s arm from the
shoulder to the tip of the index finger. Participants stood at their comfortable stance
(approximately shoulder width wide) with their toes aligned behind a fixed line marked on the
force plates to ensure that they always stood at the same position. Participants with MS wore a safety harness tethered to a rail system on the ceiling to prevent injury in case of falls.

Kinematics, kinetics and electromyograms (EMGs) were measured during the tasks. Participants wore a tight-fitting sleeveless compression shirt over their own attire for better marker placement, and were set up with the full body Vicon (Vicon Motion Systems Ltd, Oxford, UK) marker set. Participants also wore a snug fitting rubber fingertip with a smooth fabric exterior over the finger for tracking to reduce friction on the screen. Surface EMGs were recorded bilaterally from the vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), medial head of the gastrocnemius (MG), tibialis anterior (TA), and anterior deltoid (AD) from all participants. Recording electrodes (Delsys Inc, Boston, MA) were placed on the cleansed, lightly abraded skin over each muscle belly. EMG signals were sampled at 1000 Hz and bandpass filtered (25-450 Hz) prior to sampling. Kinematic data were recorded via the six motion capture cameras at 100 Hz, and kinetic data were recorded via the two force plates at 1000 Hz. All signals were recorded in Vicon Nexus software (Vicon Motion Systems Ltd).

**Experimental Protocol**

Participants were presented with two different tracking trajectories—a vertical and horizontal figure-eight and two types of tracking tasks — steady speed and ramp speed tracking. During steady speed tracking, the target would move at a constant speed at frequencies of 0.05, 0.10, 0.15, 0.20, 0.25, 0.30 or 0.35 Hz, (2-12 cycles) with each trial lasting between 30-40 seconds. During ramp speed tracking, the target would start moving at a frequency of 0.05 Hz,
and linearly increase its frequency to 0.35 Hz over a trial period of 67s. Participants would start each trial in a standing position, with their dominant arm fully extended in front of them. They would hear four short consecutive beeps to indicate that the trial was about to start, and on the fourth beep, the target would start moving. The participants were instructed to keep their index finger on the screen and track the target as closely and accurately as possible without changing their feet position. The tracking trajectory was first presented as a vertical figure-eight, and the trials of the various steady speed tracking were randomized. After every three steady speed tracking trials, participants would perform a ramp speed tracking trial. Once trials of all speeds had been completed, the tracking trajectory was switched to the horizontal figure-eight, and the same procedure was repeated. Participants were allowed to rest and take a seat in between trials. Participants completed three practice steady speed tracking trials at medium speed (0.15, 0.20 and 0.25 Hz) at the beginning of the experiment for the vertical figure-eight tracking, and again when the tracking trajectory was changed to the horizontal figure-eight.

Data Analysis

Kinematic and kinetic data were recorded and used for the reconstruction of the participant-specific full body model in Vicon. As participants made multiple continuous tracking cycles throughout each trial, the start and end of each cycle was marked as an event in Vicon, based on the location of the finger marker on the hand that they were using to track the target. The model outputs of marker trajectory, joint angles, joint moments, center-of-mass (CoM) movement, as well as the EMG data and force plate outputs were then exported for further processing in Matlab. All model outputs including marker trajectories, joint angular
range of motion, and joint moments were low pass filtered at 10 Hz using a 4th order zero phase Butterworth filter.

Each cycle of the target movement was defined as having a start point in the middle of the figure-eight, a full movement around the figure-eight, and an ending point in the middle of the figure-eight. To obtain the representative average joint kinematic profile over one tracking cycle for each steady speed trial for each participant, each cycle of ankle angle and hip angle was downsampled to a specific number of points depending on the speed of the trial (2000, 1000, 667, 500, 400, 333, 286 downsampled data points for speeds of 0.05, 0.10, 0.15, 0.20, 0.25, 0.30, 0.35 Hz respectively). These downsampled cycles were then averaged within participants to obtain the representative average joint kinematic profile of the ankle and the hip. The maximum and minimum amplitude values of the kinematic profile were calculated, and the peak-to-peak (PP) values were obtained from the difference of the maximum and minimum values.

Obtaining the representative kinematic profile over one tracking cycle for each ramp speed trial was not possible as the period of each cycle continuously changed as tracking speed increased throughout the trial. To obtain the maximum, minimum, and PP amplitudes of the kinematic data, the maximum and minimum values of each cycle were first determined throughout the trial. These values were then used to calculate the coefficients of a 3rd degree polynomial fit, which was used to estimate the maximum and minimum values at set time points in the ramp speed trial corresponding to each target speed. This process produced seven maximum and minimum values that characterized the kinematic data throughout the ramp speed trial. The PP values of the ramp speed kinematic data were then obtained from the difference between these corresponding maximum and minimum values.
The sagittal hip and ankle PP moments and PP angular range of motion were the primary parameters used as indicators of the movement strategy during tracking. To obtain a meaningful parameter to quantify the shift in hip and ankle strategy, the PP moment and angular range of motion values of the hip was divided by that of the ankle to obtain a unitless ratio of hip to ankle moment and angular range of motion. This was done for each speed for both the steady speed and ramp speed trials. This step was important as it served as a method of self-normalization within each participant’s joint kinematic amplitudes. For example, a high functioning participant might have large PP joint moments at both the hip and ankle, whereas a low functioning participant might have low PP joint moments at both the hip and ankle. Also, one participant might adopt a strategy of large movements during tracking, while another might adopt a stiffening strategy to maintain balance during tracking. This creates large variability when comparing hip to hip or ankle to ankle across participants. By taking a ratio of the PP joint kinematics, a single parameter indicative of the participant’s relative amount of hip or ankle strategy was obtained, reducing the variability during subject and group comparisons.

The center of mass (CoM) root mean square (RMS) acceleration in the anterior-posterior (A-P) and medial-lateral (M-L) directions was calculated in quiet standing and during tracking tasks to obtain a measure of balance performance. This parameter has been used in previous studies to show differences in balance performance during standing in MS (Huisinga et al., 2012b).

EMG signals were processed to obtain the timing and level of muscle activity during experimental procedures. All surface EMG recordings were notch filtered at 58-62 Hz to remove line noise and bandpass filtered at 30-200 Hz using 4th order zero phase Butterworth filters (Matlab; the Mathworks Inc., Natick, MA). For analysis, the EMG data were rectified and
low pass filtered at 5 Hz using a 4th order zero phase Butterworth filter, then ensemble averaged across cycles within each subject and trial. Analysis of the MG and TA EMGs were used to determine the amount of ankle activity during the tracking tasks.

Statistical Analysis

A repeated measures ANOVA (SPSS Inc., Chicago IL) was used to compare the group difference in effect (across seven tracking speeds of 0.05 – 0.35 Hz) on the PP joint moments, PP angular range of motion, and CoM movement, between the MS and healthy control groups. For statistical analysis of the correlation between the joint parameters and clinical scores, a linear correlation was conducted between the maximum joint moments, angles, and PP CoM movement, and the clinical scores of the Berg balance score and the 25 foot walk test. Significance was accepted at p<0.05.
Results

Kinematic Analysis

The aim of this study was to identify the hip and ankle strategies associated with standing during an arm tracking task in individuals with MS compared to healthy controls. Profiles of hip and ankle moments and angles of both groups were examined to compare the differences in the hip and ankle strategies during the tracking task. The focus was primarily on sagittal plane moments and angles; frontal plane moments and angles were also analyzed but showed no significant trends. Representative single participant plots from a single tracking trial are shown in Figure 1. Most participants’ kinematic traces reflected the cyclic characteristic of the tracking task.

Participants showed varying amplitudes of hip and ankle moments and angles, related to the different postures and tracking strategies. When each participant’s representative hip and ankle kinematic and kinetic profiles per cycle were obtained and averaged within the group (Figure 2), both groups showed similar profiles. The largest deflections in the trajectory of hip kinetics and kinematics during the vertical tracking task occurred when tracking the lower portion of the target, where bending was required to reach the target. Overall, the ankle showed smaller deflections, although plantarflexion was evident during tracking of the upper portion of the target, presumably to reach the top of the target. A similar profile was seen in both the left and right hip and ankle, as the vertical tracking task was relatively symmetric for both legs. On the other hand, the trajectories of the hip moments and angles for the horizontal tracking task were asymmetric, requiring large medial-lateral movement and weight shifting between legs throughout the task. At the ankle, plantarflexion moments corresponded to weight shifting for
the right and left sides. For both tasks, we observed that the hip moments and angles were generally larger in participants with MS, while ankle moments and angles were typically similar or smaller.

The group averages of EMG for a single speed (Figure 3) showed larger amplitudes of activation at the TA and MG muscles for the control group, compared with the MS group. EMG activation followed a typical double peak profile in each half of the cycle, likely related to the movement and weight shifting pattern inherent to the tracking task. This increase in EMG amplitude in the control group is consistent with the larger maximum moments observed in the ankle joint moment plots for the same speeds.

**Hip/ankle PP Joint Moments and Angles**

In general, there were larger relative movements and joint moments at the hip, compared to the ankle for standing, figure-eight tracking movements in participants with MS. Inspection of the group averaged kinematic and kinetic profiles for a single speed (Figure 2) suggested that the MS group had greater PP moments and angles at the hips, and generally smaller PP moments and angles at the ankles. The ratio of PP hip moments and angles relative to the PP ankle moments and angles were calculated for each group, for each test speed and are shown in Figure 4. The magnitudes of the PP hip/ankle group averaged parameters were larger in the MS group, across all tracking speeds, and for both the vertical and horizontal steady speed tracking. The difference in PP hip/ankle joint parameters was significant between groups for all tracking paradigms (Repeated Measures ANOVA p<0.05), and for 10 out of 16 parameters (Table 2).
The hip/ankle ratios of angle and moment from the ramp speed tracking showed a similar trend as the constant speed task, with the MS group showing larger hip/ankle PP joint moments and angles during both vertical and horizontal ramp speed tracking, compared with the control group. Both groups were significantly different during vertical and horizontal tracking (repeated measures ANOVA p<0.05), but significance was only detected at the right hip/ankle PP moment during vertical ramp speed tracking, and the left and right hip/ankle PP moments during horizontal ramp speed tracking (Table 2). There was also an increase in hip/ankle ratio with increasing tracking speed in both the MS and control groups, which was particularly noticeable in the left and right hip/ankle PP moment during both vertical and horizontal ramp speed tracking. This difference in hip/ankle ratio was used as an indicator of a change in balance strategy for people with MS.

The PP hip/ankle ratio was related to functional balance. For the MS group, the PP hip/ankle parameters for each subject were averaged together across all seven tracking speeds, then averaged between the left and right leg of each subject for regression with the BBS (Figure 5). Note that the controls were not included in the BBS analysis, since they all scored at the maximum value. The MS group showed a significant correlation between the PP hip/ankle moment ratio and BBS during the horizontal steady speed tracking task ($R^2 = 0.884$, p<0.001), demonstrating increasing PP hip/ankle moment ratio with increasing impairment (decreasing BBS).

Balance performance during the figure-eight tracking was similar for the MS and control groups. During steady speed tracking, the CoM RMS acceleration increased with increasing tracking speed for the MS and control group. However, the CoM RMS acceleration (A-P) was
not significantly different between groups (repeated measures ANOVA, vertical tracking p=0.513, horizontal tracking p=0.159).
Results from this study showed greater hip range of motion and joint moments than ankle during a standing arm tracking task in people with MS, compared with healthy controls.

Although both groups showed increasing ratios of hip/ankle joint moments with increasing tracking speeds, the amount of sagittal hip to ankle moment was higher in the MS group than the control group for vertical and horizontal tracking, during both steady speed and ramp speed tracking. During the tracking tasks, balance performance (measured by CoM RMS acceleration) was not significantly different between groups, or correlated to clinical measures of balance in MS. These results indicate a shift to an increased hip strategy during arm tracking involving whole body movement in people with MS, while maintaining comparable balance performance to healthy participants.

There is a shift towards a hip strategy during increasing challenges to balance. During tracking, both MS and control groups utilized an increased hip strategy with increasing perturbation size, reflected by an increasing PP hip/ankle moments of both groups with increasing tracking speed. The overall increase in moments at the hip and ankle was consistent with an increase in difficulty of the tracking task, where the participant was required to move faster to keep track of the target. It is likely that the participants utilized combined hip and ankle strategies for adaptable control of CoM in the sagittal plane (Nashner and McCollum, 1985) in different magnitudes and temporal relations (Horak and Nashner, 1986) to optimally maintain balance during tracking. The ankle strategy dominated during low speed tracking as the ankle
strategy repositioned the CoM by moving the entire body as a single segmented inverted pendulum (Kuo and Zajac, 1993) through the production of ankle torque, suitable for correcting small amounts of sway resulting from slow, small perturbations and maintaining vertical posture while moving the CoM (Kuo, 1995). As the tracking speed increased, there was a need for an increased hip strategy, which moved the body as a double segment inverted pendulum (Runge et al., 1999), with counter phase motions at the ankles and hip, suitable for quick postural adjustments for correction of larger, more rapid perturbations. This shift towards a hip strategy during increasing external perturbations has been observed in healthy individuals (Liaw et al., 2009), and it is likely that both groups utilized this strategy when the tracking speed increased.

There was a larger shift towards a hip strategy in MS than healthy participants. Although the shift towards a hip strategy with increasing tracking speed was observed in both groups, the participants with MS generally used the hip strategy more than the controls. This was reflected by the consistently larger PP hip/ankle moments and angles in the MS group compared with the control group for all tracking speeds. This shift towards a hip strategy in the MS group is consistent with studies that report an increased hip mechanism during external perturbations in neurologically impaired populations (Termoz et al., 2008; Hwang et al., 2009). The ankle strategy typically dominates during normal stance and comfortable balance tasks, whereas a narrow stance width or increased balance difficulty requires changes to the whole body movement pattern (Danis et al., 1998) and decreases the role of the ankle while increasing the role of the hip (Gatev et al., 1999). This shift in strategy away from the ankle might be explained by limits of ankle muscles, which are largely responsible for phasic control of anterior-posterior balance during quiet standing (Borg et al., 2007). However, the ankle strategy requires more muscle activation (based on a biomechanical model with activation normalized by muscle peak
isometric force) than the hip strategy with increased perturbation size, and a combination of ankle and hip movement with an increased ratio towards the hip, is favored with increasing speed (Kuo and Zajac, 1993). Hence, deficits at the ankle in people with MS might be responsible for the baseline shift towards the hip strategy during challenging perturbations.

Interaction between Movement Strategy and Balance

Although the MS group demonstrated reduced balance function, measured clinically with the BBS and kinematically during quiet standing, the balance performance during tracking was not significantly different between the MS and control groups. This was confirmed using the CoM RMS acceleration, identified as a possible primary sensory input for maintaining balance (Nataraj et al., 2012a, 2012b), which indicated that the MS group was able to sufficiently maintain comparable balance performance to the control group during tracking. Ataxic-spastic participants with MS are able to compensate for proprioceptive deficits with more efficient control strategies during standing to achieve the same postural performance as spastic participants with MS (Rougier et al., 2007). Furthermore, minimally impaired adults with MS are able to use an adapted reaching strategy that allows them to stay within their reduced limits of stability to perform the same reaching task as healthy controls (Karst et al., 2005). Hence, it appeared that participants with MS in the current study were able to compensate for balance deficits and maintain balance performance during tracking.

The relationship between the amount of hip strategy used and balance impairment in MS is unclear. Although the PP hip/ankle moment ratio was strongly correlated to BBS during horizontal tracking, no clear regression was observed during vertical tracking. It is possible that
the larger size of the horizontal tracking task required a greater range of whole body movement, resulting in balance deficits driving the use of the hip strategy in MS. However, without a similar trend during vertical tracking, there is insufficient evidence to implicate balance impairment as a driving force for a shift towards the hip strategy in MS.

Causes of Changes to Movement Strategy

Deficits in sensory inputs and/or errors in motor commands in people with MS might contribute to an increased hip strategy. In MS, proprioceptive sensory feedback signals might be erroneous or delayed, causing a decreased use of proprioceptive information (Feys et al., 2006). Additionally, prolongation of spinal cord somatosensory conduction can cause delays in postural response onset (Cameron et al., 2008), indicating that people with MS receive delayed somatosensory information, and subsequently respond slower to perturbations. As multiple, redundant local sensory signals are integrated as part of a hierarchical feedback control system (Ting, 2007), a decrease in one aspect of sensory input might increase reliance on other sensory signals. Additionally, damage to descending motor pathways might result in inaccurate motor commands that are compensated later through additional sensory-based corrections (Casadio et al., 2008) and subtle alterations in sensorimotor control (Solaro et al., 2007). As functional performance might be correlated to the number of reliable sensory inputs (Cattaneo and Jonsdottir, 2009), sensory deficits in MS are likely to cause a change in strategy. Studies of postural strategies used by patients with sensory loss have shown that both biomechanical constraints and available sensory information are important to strategy selection. For example, patients with complete vestibular loss are unable to utilize a hip strategy for efficient control of
equilibrium and balance recovery (Horak et al., 1990). However, impaired somatosensory information from the lower limbs results in an inability to use an ankle strategy effectively, and increased reliance on a hip or stepping strategy (Horak et al., 1990; Runge et al., 1994). Hence, it is likely that the participants with MS in this study utilized an increased hip strategy due to sensorimotor impairments.

The ability to correctly anticipate a postural change or perturbation might be impaired in MS, resulting in less automated and insufficient postural control during challenging perturbations, similar to observations in elderly (Laessoe and Voigt, 2008). In this study, we used an internal perturbation in the form of a continuous tracking task, presenting a certain amount of predictability in the target. As this was a standing arm tracking task, the continuous arm movement likely involved anticipatory non-focal muscle activity that transported the arm and resisted the perturbation caused by the arm movement to minimize the impact of predictable perturbations to balance (Tyler and Karst, 2004). This movement also required dynamic changes in the trunk and lower extremities to stabilize the body and initiate and assist whole body reaching (Kaminski and Simpkins, 2001). The participants with MS might have a diminished ability to produce directional specific patterns of anticipatory muscle activation and postural shifts associated with a rapid arm movement, resulting in reduced balance function (Krishnan et al., 2012) and requiring an increased hip strategy. The inability to anticipate postural perturbations might be related to weakness of muscles associated with balance recovery (Carty et al., 2012), as well as asymmetries in lower limb power (Chung et al., 2008). Hence, the inability to make predictive postural changes might diminish balance performance in MS, prompting the use of an increased hip strategy.
Measurements of hip compensatory strategies might provide estimates of impairment that are not obtainable from functional measures. Currently, the Expanded Disability Status Scale (EDSS) and Multiple Sclerosis Functional Composite (MSFC) are used for assessing MS impairment, particularly for clinical trials. These scales are multidimensional and place an emphasis on ambulation and balance function. Several clinical measures of balance like the Berg Balance Scale, Timed 25-Foot Walk, Six-minute Walk, and Timed Up and Go, have been used to reliably quantify the level of balance impairment in moderately impaired study participants, but might not be as useful in a heterogeneous or more impaired clinical population (Learmonth et al., 2012). Additionally, the functional reach test, which has been used to distinguish between MS individuals with poor balance (Frzovic et al., 2000), might be a weak measure of stability limits as it does not take into account compensatory mechanisms (Jonsson et al., 2003). As the compensatory hip strategies employed by people with MS might vary with impairment, clinical assessments of walking and balance become less accurate in assessing the underlying degree of functional impairment. The protocol in the current study distinguished the difference in hip versus ankle strategy between individuals with MS and healthy controls in the context of a voluntary movement, which might be a useful measure of pathophysiology in MS.

An improved understanding of the interactions between hip versus ankle strategy and balance function might help to direct rehabilitation interventions in MS. There is currently no widely used standard of prescribed rehabilitation that improves balance in people with MS. Often, AFOs or walking aids are prescribed to assist with walking and balance, and to prevent falls and accidents during ambulation. The use of these devices inhibits access to ankle strategies, which could impact compensatory strategies for balance. Balance rehabilitation in
different sensory contexts is useful in reducing the rate of falls and improving dynamic balance in people with MS (Cattaneo et al., 2007). For example, asymmetrical placement of a light weight on the torso assists MS patients in maintaining balance during static and dynamic activities (Gibson-Horn, 2008; Widener et al., 2009). Training interventions focusing on mechanisms related to dynamic stability have also been shown to increase ankle strategy (Lindemann et al., 2012), and strength training reduces center of pressure variability in people with MS (Huisinga et al., 2012a). Hence, by understanding how compensatory strategies affect performance during functional movements, unique rehabilitation interventions can be used to either strengthen the underlying muscles, or bolster the effectiveness of the compensatory strategy.

In conclusion, the observation that the PP hip/ankle ratio of joint moments was significantly larger in the MS group than the control group suggests that people with MS increase their hip strategy and reduce their ankle strategy during standing arm tracking compared to age-matched controls. This adaptation is likely due to increased sensorimotor impairment at the ankle, but it allows people with MS to achieve a similar task performance as controls.

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### Table 1 Clinical Features of Participants with MS

<table>
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<tr>
<th>Subject</th>
<th>Age (yrs)</th>
<th>Years with MS</th>
<th>9HPT (s)</th>
<th>SDMT (of 7)</th>
<th>FSS (of 7)</th>
<th>BBS (of 56)</th>
<th>25FWT (s)</th>
<th>Modified Ashworth Score*</th>
<th>Medications</th>
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<td>21.30</td>
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<td>56</td>
<td>6.52</td>
<td>6 Baclofen, Ampyra</td>
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<td>6</td>
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<td>18.40</td>
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<td>20.52</td>
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<td>4.67</td>
<td>52</td>
<td>7.62</td>
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</table>

Clinical tests include the Nine Hole Peg Test (9HPT), Symbol Digit Modalities Test (SDMT), Fatigue Severity Scale (FSS), Berg Balance Scale (BBS), and 25 Feet Walk Test (25FWT).

* Modified Ashworth Score is calculated by summing the individual Modified Ashworth Scale muscle scores (0-4) from the left and right knee extensors, knee flexors, ankle plantarflexors, ankle dorsiflexors (total of 8 muscles), to obtain a scoring range of 0 (low muscle tone) to 32 (high muscle tone).
Table 2 Significance of PP Hip/ankle Parameters

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<thead>
<tr>
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<th>Vertical SS</th>
<th>Horizontal SS</th>
<th>Vertical RS</th>
<th>Horizontal RS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Combined</td>
<td>0.006*</td>
<td>0.000*</td>
<td>0.009*</td>
<td>0.013*</td>
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<tr>
<td>Left hip/ankle PP moment</td>
<td>0.011*</td>
<td>0.003*</td>
<td>0.088</td>
<td>0.007*</td>
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<td>0.004*</td>
<td>0.000*</td>
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<tr>
<td>Left hip/ankle PP angle</td>
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<td>0.016*</td>
<td>0.252</td>
<td>0.346</td>
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<td>0.030*</td>
<td>0.029*</td>
<td>0.077</td>
<td>0.063</td>
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</table>

p-values for vertical and horizontal steady speed (SS) and ramp speed (RS) tracking tested with a repeated measures ANOVA. The Combined analysis compared all four parameters between the MS and control group. Significance (p<0.05) is marked with a *. 
Figure Legends:

**Figure 1 Representative Single Participant Data.** Representative kinematic, kinetic and EMG data from a representative control and a representative MS participant during a vertical 0.25 Hz tracking trial. At the hip, flexion angles and external moments are positive, extension angles and external moments are negative. At the ankle, dorsiflexion angles and external moments are positive, plantarflexion angles and external moments are negative.

**Figure 2 Group Average Kinematic and Kinetic Plots for Figure-Eight Tracking.** MS (solid line) and control (dashed line) group averages of hip and ankle sagittal plane moments and sagittal plane angles during a 0.20 Hz vertical SS tracking task. Each plot represents a full cycle of tracking (0-100% cycle) around the figure-eight trajectory. Positive moments indicate hip extension or ankle plantarflexion moments, and negative moments indicate hip flexion or ankle dorsiflexion moments. Positive angles indicate hip flexion or ankle dorsiflexion, and negative angles indicate hip extension or ankle plantarflexion. The MS group was observed to have increased hip moment and angular range of motion and reduced ankle moment and angular range of motion compared with the control group.

**Figure 3 Group Average EMG Plots.** MS (solid line) and control (dashed line) group averages of the left and right leg MG and TA muscles during a 0.35 Hz vertical and horizontal steady speed (SS) tracking task. Each plot represents a full cycle of tracking (0-100% cycle) around the figure-eight trajectory. The MS group was observed to have reduced MG and TA activity compared with the control group.

**Figure 4 PP Hip/Ankle Ratio of Joint Kinematics and Kinetics for each Speed.** Group averages of PP hip/ankle ratio of joint moments and joint angles for each steady speed and ramp
speed vertical and horizontal tracking. The MS group showed an increased PP hip/ankle ratio for joint moments and angles across all tracking speeds, compared with the control group. * Indicates significance across pairwise comparisons between same speeds (p<0.05). Standard deviation bars are indicated for each speed (Negative bars for control group and positive bars for MS group).

Figure 5 Steady Speed PP Hip/ankle Ratios Averaged Across All Speeds compared with BBS. The participants with MS showed a strong regression between PP hip/ankle moment ratio and BBS during steady speed horizontal tracking, demonstrating that the PP hip/ankle moment ratio increased with decreasing BBS. * indicates regression significance (p<0.05).
Figure 1
Figure 2
Figure 3

0.35 Hz SS Vertical Tracking

0.35 Hz SS Horizontal Tracking
Figure 4
Figure 5
REFERENCES


0.35 Hz SS Vertical Tracking

- **Left MG**
- **Right MG**
- **Left TA**
- **Right TA**

0.35 Hz SS Horizontal Tracking

- **Left MG**
- **Right MG**
- **Left TA**
- **Right TA**