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Multiple Sclerosis Alters the Mechanical Work Performed on the Body's Center of Mass During Gait

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Patients with multiple sclerosis (MS) have less-coordinated movements of the center of mass resulting in greater mechanical work. The purpose of this study was to quantify the work performed on the body's center of mass by patients with MS. It was hypothesized that patients with MS would perform greater negative work during initial double support and less positive work in terminal double support. Results revealed that patients with MS perform less negative work in single support and early terminal double support and less positive work in the terminal double support period. However, summed over the entire stance phase, patients with MS and healthy controls performed similar amounts of positive and negative work on the body's center of mass. The altered work throughout different periods in the stance phase may be indicative of a failure to capitalize on passive elastic energy mechanisms and increased reliance upon more active work generation to sustain gait.

Keywords: dynamic walker, external work, energy, biomechanics, locomotion

To describe human locomotion, an idealized double pendulum model has been used.[1,2] In this model, the body's center of mass rotates at the end of an inverted pendulum before it reaches the end of its arc motion and transitions to the other leg, which proceeds to act as another inverted pendulum. As the inverted pendulum reaches the end of its arc and the body's center of mass shifts to the other leg, the leg changes from an inverted pendulum to a suspended pendulum and the leg swings forward, thereby positioning itself for the next inverted cycle. This motion allows efficient transfer of the energy of the system between potential and kinetic forms.[2,3] In this idealized model, the maintenance of mechanical energy requires only minimal energy from gravity to overcome step-to-step transitions. In a real-world walking scenario, the energy exchange from potential to kinetic and vice versa is not perfect: energy is lost to various entities such as friction, heat, sound, as well as step-to-step transitions. However, the inverted pendulum model is closely mimicked in passive walking robots that only require gravitational potential energy to overcome such natural energy losses as they descend down a mild slope.[4–6] In these robots, there needs to be a careful balance between the gravitational potential energy input to the system and the energy that is lost because excessive or insufficient energy will cause the passive walker to fall over.

Humans commonly negotiate various inclines and declines as well as modulate their speed while walking. Thus, without the dependence on gravitational potential energy as well as a need to modulate speed, humans rely on muscles to produce force and contribute energy to the system. By modulating the timing of muscle firing throughout the lower limb, humans are able to sustain bipedal locomotion. As Saunders et al[1] first described with the six determinants of gait, the ultimate effect of coordinated walking is the smooth motion of the body's center of mass as it moves through space on a sinusoidal path. It would seem that with numerous variables (eg, muscle size, muscle type, limb size) that can affect every step, the neuromusculoskeletal system is able to create a highly functional amalgamated whole to efficiently maintain a gait pattern.[2]

In light of such coordinated movement, it is intriguing to consider a compromised neuromuscular system that is not able to operate in such an efficient pattern. Multiple sclerosis (MS) is a neurological disease that results in progressive demyelination of axons followed by dendritic scarring that prevents repair of the damaged axons.[7] MS patients live with symptoms such as limb weakness, gait ataxia, depression, vertigo, and other central nervous system issues.[7,8] In addition, up to 90% of patients with MS will experience spasticity.[9] Within 10–15 years of

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disease diagnosis, up to 80% of MS patients report gait problems due to muscle weakness or spasticity, fatigue, or balance impairments.[10] Patients with MS typically walk slower, have shorter stride lengths, spend more time in double stance, and have reduced high-frequency content within their vertical ground reaction forces during walking,[11–13] which results in a notable gait apraxia.[14] Patients with MS walk with increased metabolic energy demands[15,16] and increased oxygen cost compared with healthy controls.[17] The reasons for these excessive metabolic demands during walking are not entirely clear. Despite these problems, a large percentage of patients are still ambulatory and highly functional. Gait abnormalities due to altered mechanics during walking in patients with MS are likely contributors to the increased metabolic cost. With altered gait mechanics, the body's center of mass would move along a path atypical from that described by Saunders et al[1] and require increased mechanical energy to sustain the movement of the center of mass during gait.

Measurement of work performed on the body's center of mass examines the change in energy of the body's center of mass as it travels from point to point. The path of the body's center of mass is accounted for while also relating the change in mechanical energy throughout the gait cycle. This is done by factoring in the force (ie, ground reaction force) that is displacing the body's center of mass. The work performed on the body's center of mass has been used similarly to examine gait in children with cerebral palsy.[18] Cerebral palsy also affects the central nervous system. Similar to MS patients, cerebral palsy patients experience lower extremity spasticity, although to a much greater severity. Kurz et al[18] found children with cerebral palsy performed more negative work with the lead leg and less positive work with the trailing leg during the transition between limbs. Positive work indicates energy produced whereas negative work indicates energy being dissipated or stored as potential energy. Kurz et al[18] also reported that children with cerebral palsy performed increased positive work on the body's center of mass during single support to maintain locomotion. Shifting positive work to the single support phase is less metabolically efficient than generating energy during push-off with the trailing leg.[19,20] This adaptation was reported as a likely contributor to the increased metabolic cost in cerebral palsy gait.[18] Similar to Kurz et al's study on cerebral palsy patients, investigating the work performed on the body's center of mass in patients with MS may reveal if and how these patients are adjusting their mechanical energy being performed during walking to get from point to point.

Thus the primary objective of this study was to investigate the mechanical work performed on the body's center of mass during walking in patients with MS as compared with healthy controls. Based on the findings from Kurz et al's study on patients with cerebral palsy,[18] it was hypothesized that MS patients would similarly perform greater negative work on the body's center of mass during initial double support and less positive work in terminal double support as compared with healthy controls. Furthermore, we expected decreased negative work and increased positive work during single support to compensate for changes during double support phases. As a secondary objective, we investigated the corresponding average powers during stance.

Methods

Subjects

Nineteen patients (Table 1) diagnosed with MS (age 42.9 ± 11.0 y) and 19 healthy controls (age 39.3 ± 10.7 y) were recruited for participation in this study. Patients and controls were matched according to self-selected walking speeds. All participants provided informed consent in accordance with the University of Nebraska Medical Center's Institutional Review Board. Inclusion criteria included cognitive ability to give informed consent. For MS patients, they were also required to have an Expanded Disability Status Score (EDSS)[21] score of 1.0–6.0 and have physical and neurological examinations that were "clinically acceptable," where evidence is required that the MS patient's physical and neurological conditions would not place the patient at unnecessary risk. All MS patients were assessed by an MS care specialist (MF).

Experimental Design and Procedures

All data collections took place at the Nebraska Biomechanics Core Facility. Participants wore a tight fitting spandex uniform and athletic shoes. Retroreflective markers were placed at the sacrum, heel, and top of the second metatarsal phalangeal joint. Participants walked across a 10 m walkway with an embedded force platform (Kistler 9281B, Kistler Instrumentation Corporation, Amherst, NY) collecting ground reaction forces at 600 Hz. Three-dimensional marker positions were recorded in real time with an 8-camera motion capture system (Motion Analysis, Santa Rosa, CA, USA) sampling at 60 Hz. The two legs of each participant were tested separately in random order. Five successful trials were collected for each leg. A successful trial occurred when only the leg of interest had a single step contact the force platform and landed entirely within the perimeter of the force platform. In

Table 1 Demographics, mean ± SD, for healthy controls and patients with multiple sclerosis

	Healthy Control (n = 19)	Multiple Sclerosis (n = 19)
Age (years)	39.3 ± 10.7	42.9 ± 11.0
Height (cm)	172.0 ± 8.9	169.9 ± 9.4
Mass (kg)	72.7 ± 14.4	86.3 ± 17.8*
Velocity (m/s)	1.22 ± 0.25	1.17 ± 0.20
Cadence (step/s)	1.96 ± 0.16	1.99 ± 0.17
EDSS	n/a	2.97 ± 1.53

*Significant difference from healthy controls at $P < .05$.

between each trial, participants were required to take 1 min of rest to prevent fatigue. All participants walked at their self-selected comfortable speed.

Data Analysis

Work performed on the body's center of mass was calculated as the integral of the dot product of the ground reaction force and velocity of the body's center of mass during the stance phase of the limb in contact with the force platform.[19,20] Stance phase was divided into 3 separate periods:[22] initial double support, single support, and terminal double support (Figure 1). Initial double support coincided with the time the foot came in contact with the force platform while the contralateral limb was still in contact with the floor. During single support, the contralateral limb is in swing phase. The terminal double support was the period when the foot was still in contact with the ground and the contralateral foot came into contact with the ground. These time periods within the stance phase were determined through heel and toe marker kinematic data using techniques described by O'Connor et al.[23] The velocity of the body's center of mass in three orthogonal directions was estimated as the time derivative of the sacral marker position. The sacral marker can serve as an accurate estimate for the body's center of mass in individuals ambulating at speeds below 1.4 m/s, as speeds increase from 1.4 m/s there is a divergence from the sacral marker position and the body's center of mass position as found from the double integration of ground reaction forces.[24] It was necessary to use the sacral marker for the body's center of mass as only one force platform was available.

For each participant, right and left leg trials were averaged to get a representative step. For each leg, the periods of initial double support, single support, and terminal double support were analyzed for positive and negative work. Positive work occurs when the force from the leg occurs at an acute angle with velocity vector of the center of mass, resulting in a positive dot product.[25] Negative work occurs when the force from the leg acts at an obtuse angle with the velocity vector of the center of mass, yielding a negative dot product.[25] The total positive and negative work over the entire stance phase was calculated for each leg. Work was normalized to each patient's body mass.[26] All calculations and normalizations were done through custom software in Matlab (Matlab 2007, Mathworks Inc., Concord, MA, USA). The variables of interest were as follows (Figure 2): positive work initial double support, negative work initial double support, positive work single support, negative work single support, positive work terminal double support, negative work terminal double support, total positive work, and total negative work.

The average power was calculated for each corresponding subphase of stance that work was calculated.

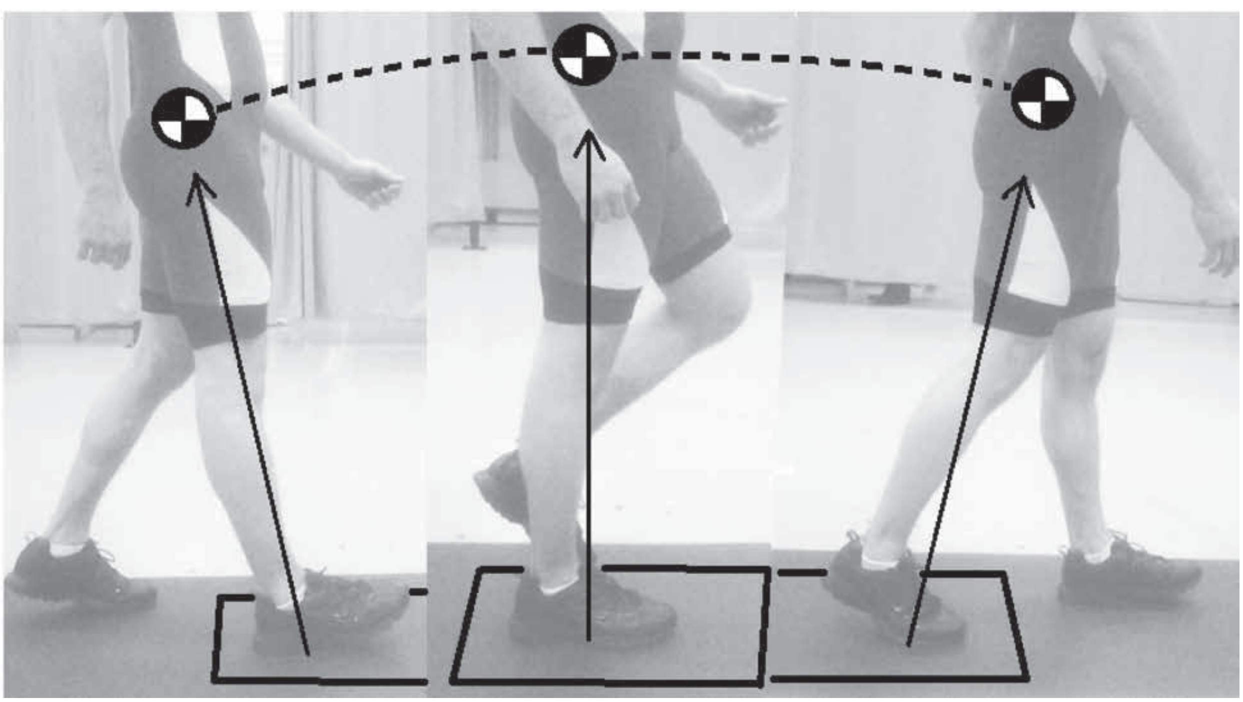

Figure 1 — For this image, the right leg is of interest. Initial double support occurs when the right leg leads the trailing left leg and both feet are in contact with the ground. Single support is when only the right leg is on the ground while the left leg is in swing phase. Terminal double support is occurring when the right leg is the lagging leg and both feet are in contact with the ground. Note the arc-like path of the body's center of mass with its approximate location in each of the three periods. Arrows are showing direction of ground reaction force.

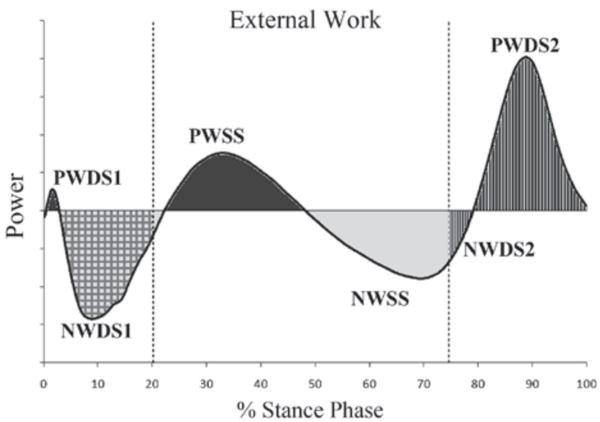

Figure 2 — Example of work performed on the body's center of mass. Positive work occurs above the abscissa, negative work occurs below the abscissa. Vertical dashed lines separate initial double support (hatched area), single support (solid area), and terminal double support (striped area). Positive work initial double support (PWDS1); negative work initial double support (NWDS1); positive work single support (PWSS); negative work single support (NWSS); positive work terminal double support (PWDS2); negative work terminal double support (NWDS2).

This calculation was performed by multiplying the calculated work in each subphase with the individual's average step frequency. Each participant's average step frequency was calculated as the average of the inverse of the step time for each trial. Group means were calculated for these variables across participants. Differences for each variable between MS patients and healthy controls were tested with independent *t* tests. Significance was set at the .05 level. For those comparisons that were significant, effect sizes were calculated with Cohen's *d*.[27] Cohen defined values of $d = 0.2$ as a small effect, $d = 0.5$ as a medium effect, and $d = 0.8$ as a large effect.[27]

Results

Compared with healthy controls, MS patients ambulated with less positive work in terminal double support when the leg is in a trailing position ($P = .003$, $d = 1.069$; Figure 3C). MS patients walked with less absolute amounts of negative work in single support ($P = .036$, $d = 0.727$; Figure 3B) but more in terminal double support ($P = .006$, $d = 1.015$; Figure 3C). There were no other significant differences for work (positive work initial double support: $P = .932$; negative work initial double support: $P = .807$; positive work single support: $P = .307$; total positive work: $P = .248$; total negative work: $P = .342$; Figure 3D).

When we examined the average power per step of the MS patients, only the positive power ($P = .014$, $d = 0.860$) and the negative power ($P = .004$, $d = 1.070$; Figure 4C) in terminal double support were significantly less than healthy controls. The average power for all other subphases in stance was not statistically different (positive power initial double support: $P = .932$; negative power initial double support: $P = .807$; positive power single support: $P = .307$; negative power single support: $P = .307$; total positive power: $P = .248$; total negative power: $P = .342$; Figure 4).

Discussion

MS patients walk with an altered work profile throughout the stance phase but perform similar amounts of mechanical work to move from point to point. We measured

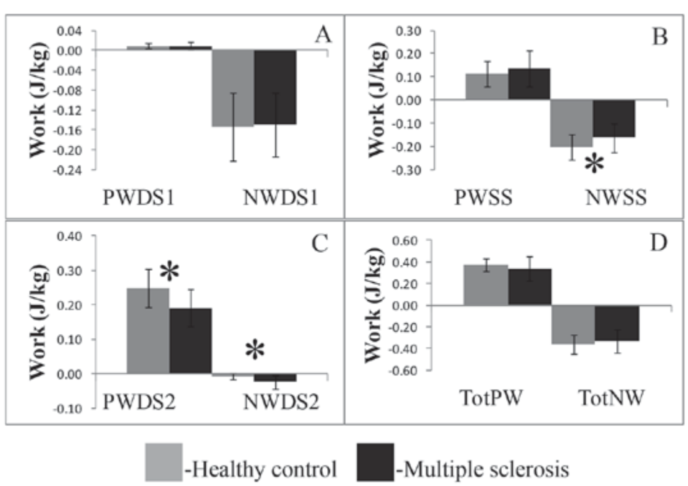

Figure 3 — Group means for work performed on the body's center of mass at different phases for a single step. (A) No differences were found in initial double support, (B) MS patients performed less negative work in single support, (C) MS patients performed less positive work and more negative work in terminal double support when push-off is occurring, and (D) no differences were found for the summed entire stance phase. Positive work initial double support (PWDS1); negative work initial double support (NWDS1); positive work single support (PWSS); negative work single support (NWSS); positive work terminal double support (PWDS2); negative work terminal double support (NWDS2); total positive work (TotPW); total negative work (TotNW). *Significant at $P < .05$.

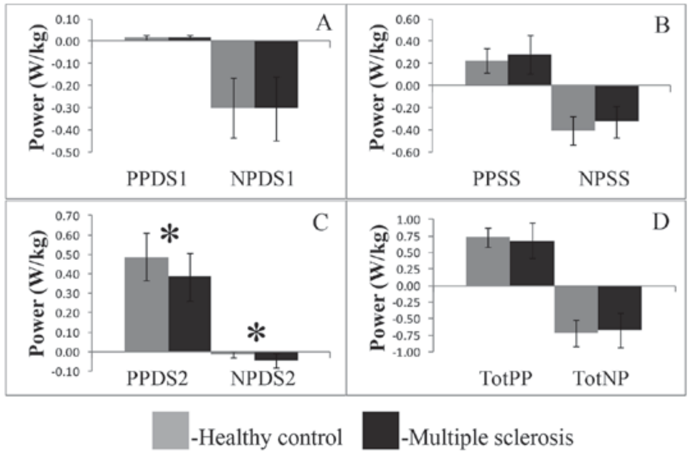

Figure 4 — Group means for average power performed on body's center of mass at different phases. (A,B) No differences were found in initial double support or single support. (C) MS patients performed less positive work and more negative work in terminal double support when push-off is occurring. (D) No differences were found for the entire stance phase. Positive power initial double support (PPDS1); negative power initial double support (NPDS1); positive power single support (PPSS); negative power single support (NPSS); positive power terminal double support (PPDS2); negative power terminal double support (NPDS2); total positive power (TotPP); total negative power (TotNP). *Significant at $P < .05$.

altered amounts of positive and negative work during single and terminal double support in patients with MS. This finding partially supported our hypothesis.

Model simulations have shown that the most mechanically efficient method of walking requires minimal work to be performed during single support. The bulk of the positive work occurs in terminal double support when the leg is in a state commonly referred to as "push-off."[19,25] MS patients perform less positive work in terminal double support, consistent with findings of reduced ankle power generation in terminal stance previously reported in MS patients.[28] The majority of negative work in single support and terminal double support would occur continuously in the stance phase and before the large positive power burst at the end of stance (Figure 2). This period has been referred to as the "pre-load" phase.[25] The negative work in preload is largely associated with elastic energy storage in the Achilles tendon.[25] This negative work slows the velocity of the body's center of mass as it progresses into the terminal double support. A diminished amount of negative work during single support would seem to indicate decreased elastic energy storage within the Achilles tendon as well as a quicker return to double support. The slowed progression of the body's center of mass would now occur with the contralateral foot on the ground. This sort of deviation would correspond with previous findings of increased double support time in MS gait[11,12] and appeared to be consistent in our patients with MS (Figure 5). In the case of MS patients, they do not seem to be allowing adequate time to perform negative work in single support. They quickly progress back to a double support period, an inherently more stable position in the gait cycle. However, this adaptation could be reducing the stored elastic potential energy in the Achilles tendon. The increased amount of negative work in double support is then slowing the body's center of mass at a time when it should be primarily producing positive work. As a result, MS patients likely have diminished contribution of stored elastic potential energy in the Achilles tendon to contribute to push-off.

Sawicki and Ferris[29] have used exoskeletons providing increased ankle power to determine that between 44% to 84% of all push-off work at the ankle is recovered from stored elastic energy within the Achilles tendon. Based on this, if MS patients are storing a decreased amount of elastic energy in the Achilles tendon, they would need increased reliance on active power generation to maintain locomotion. This would result in increased metabolic cost. In this scenario it would seem that MS patients are choosing a more stable position at the expense of an increased metabolic cost. Motl et al[17] reported values of 0.202 ± 0.023 mL·kg^{-1}·m^{-1} for MS patients compared with 0.186 ± 0.010 mL·kg^{-1}·m^{-1} for healthy controls ambulating at 0.9 m·s^{-1}. Olgiati et al[15] reported a greater discrepancy of 0.267 ± 0.018 mL·kg^{-1}·m^{-1} for MS patients compared with 0.162 ± 0.008 mL·kg^{-1}·m^{-1} for healthy controls. Olgiati et al[15] attempted to relate the increased cost of walking to spasticity; however, they concluded that only 40% of the variance could be explained by spasticity. Such conclusions would then necessarily mean that there is still 60% unaccounted variance, which leaves a very plausible scenario that a percentage of the greater metabolic cost of walking is related to the altered mechanics occurring independent of spasticity. These altered mechanics are resulting in changes in the work performed on the body's center of mass. Despite the changes in the work performed at different periods of the stance phase, the overall total positive and negative work through the entire stance phase were similar. The similar amounts of total positive and negative work are expected since

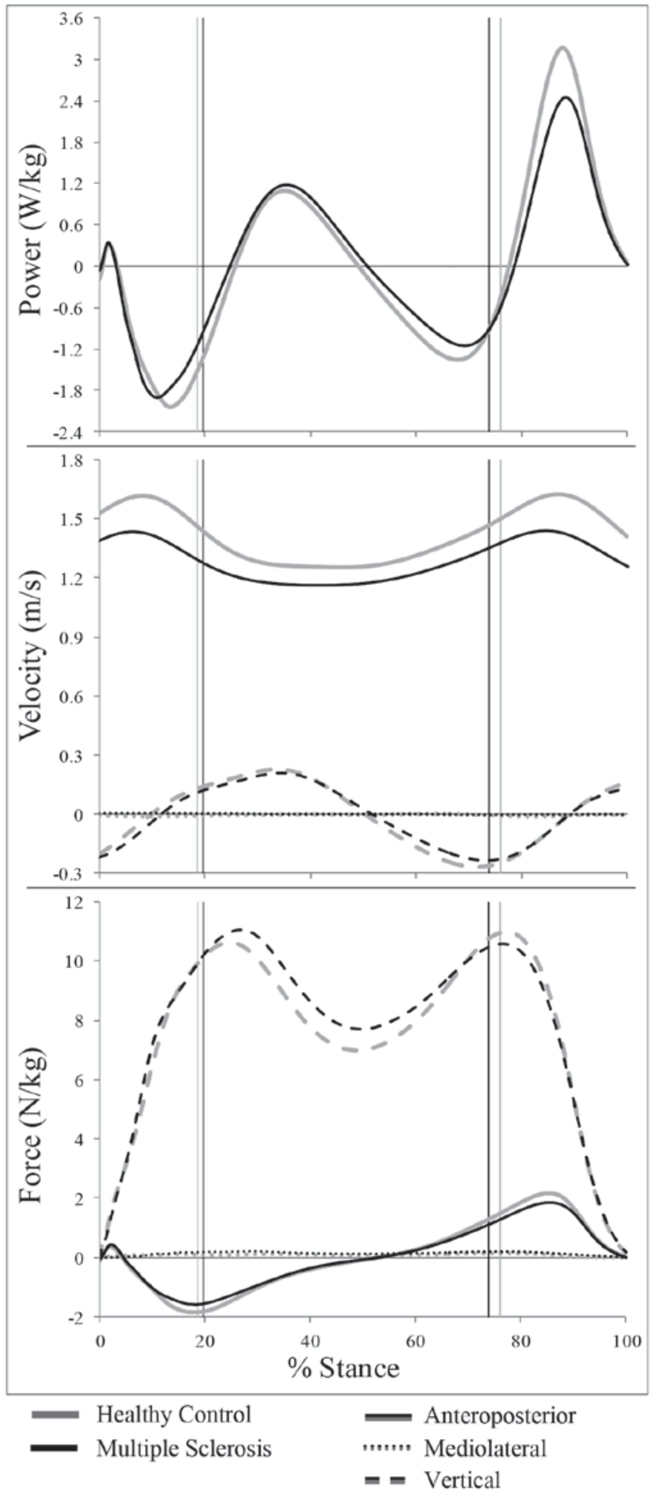

Figure 5 — Group mean ensemble curves for instantaneous work (top), center of mass velocity (middle), and force (bottom). Curves were generated by interpolating all trials to 101 points, then averaging values across each point for all trials. The group differences in the external work seem to reflect differences in velocity and force. Patients with MS seem to have decreased peak forces during braking (anteroposterior) and propulsion (anteroposterior and vertical). Patients with MS also seemed to ambulate at slower velocity despite no statistical difference. Vertical lines mark the start and end of single support for healthy controls (gray) and patients with MS (black). Note that work and power calculations were performed in real time and not normalized stance time.

participants ambulated with similar average velocities through the collection walkway.

To further elucidate the above relationships, we also calculated the average work rate. The average power was calculated by multiplying work by the average step frequency. If patients with MS ambulate with increased metabolic power[17] but similar mechanical work, then perhaps the external mechanical power will be elevated. However, our findings for total positive power and negative power were similar between groups. The differences between patients with MS and healthy controls during terminal double support remained significant (Figure 4). The average power findings only seem to further highlight major deficits during the critical phase of push-off in patients with MS.

Examination of the mean ensemble force and velocity curves for each group provides additional insight into the reasoning for the altered work performed on the body's center of mass (Figure 5). The difference in anteroposterior and vertical forces during terminal double support may be a contributor as patients with MS appeared to have decreased peak forces. Furthermore, while the statistical comparison for group velocities showed no difference, the mean ensemble curve for anteroposterior velocity would seem to indicate that the healthy controls walked faster. This difference between the mean ensemble curve and the group mean velocities may be due to the process for the mean ensemble curve generation. Curves were generated by interpolating all trials to 101 points, and then averaging all trials across these points. Group velocities were compared by calculating each individual's average velocity. Thus, our process of matching self-selected walking velocity for patients with MS to healthy controls may not entirely remove the effect of walking velocity. In addition, it seems that the decreased negative work in single support may be the result of both diminished peak power and less time in single support. Finally, inspection of the mean ensemble force and velocity curves shows the minimal contribution of the mediolateral direction in the work performed on the body's center of mass.

There are limitations to this study. First, the individual limbs method for calculating work performed on the body's center of mass as described by Donelan et al[20] utilizes a method of integration of ground reaction forces to derive the center of mass velocity. The use of a single force platform prohibits this technique and as such the sacral marker was used to estimate the body's center of mass motion. This approach has shown to be an accurate estimation for the vertical location of the body's center of mass in gait under 1.4 m/s.[24] All but 3 MS patients and 7 healthy controls ambulated at velocities below 1.4 m/s. Since only a minority of the participants ambulated above 1.4 m/s, with the greatest individual average velocity being 1.48 m/s, we are confident that although this may have slightly affected the magnitudes of values, the overall effect of decreased negative work and positive work in MS patients would persist even with the use of dual force platforms. Second, the participants were velocity matched and a *t* test found no statistical difference in walking velocity (P = .496). However, future research with fixed, standard speeds for all participants should be considered to accurately account for velocity. We chose not to have subjects walk at a standard speed as this may have altered their natural walking pattern. However, the mean ensemble anteroposterior velocity curve seems to illustrate the potential for velocity to still play a role in the differences found in work and power. Healthy individuals typically have a faster self-selected speed. Matching healthy controls with MS patients resulted in comparison with the fastest MS patients. As a result, there may be adaptations that these "faster" MS patients are able to use when ambulating. It is possible that an analysis of work performed on the body's center of mass for "slower" MS patients would show either a more dramatic effect of that measured in these participants, or possibly further shifts in the work at other points in the stance phase than just those measured. Future studies should consider an approach to compare those MS patients that are "faster" and those that are "slower," possibly revealing different mechanisms used to maintain a faster self-selected speed. Furthermore, in our study we did not consider measures of spasticity for our subjects, making it difficult to determine the degree of altered mechanics due to spasticity. Future research should examine the relationship between spasticity and work performed on the body's center of mass in patients with MS. Finally, we have chosen to analyze the work performed on the body's center of mass as this provides a clinically meaningful measure of the amount of mechanical energy that is responsible for moving the body from point to point during walking. Other forms of mechanical energy such as internal work or joint work may be able to provide further information regarding mechanical energy during walking in MS patients. Specifically, work performed on the body's center of mass underestimates mechanical work in relation to metabolic cost[30] as it fails to account for co-contractions. Future studies should consider joint work in combination with electromyography to provide insight into the amount of mechanical energy and co-contractions performed at each joint. These measures combined with work performed on the body's center of mass can provide further detail on the inefficiency of walking in MS patients.

In conclusion, MS patients ambulate with altered patterns of work performed on the body's center of mass. They perform less positive work in terminal double support combined with less negative work in single support through each step. In light of previous findings of increased metabolic cost of ambulation for MS patients,[15,17] it is possible that the decreased negative work in single support is leading to less passive, elastic energy storage. With reduced stored elastic energy subsequently being released during push-off in late stance, there is less positive work in terminal double support. These altered mechanics may reflect a desire for increased stability via increased double support time. MS patients seem to be sacrificing energetics for mechanical stability. However, further research is needed combining electromyography and work analyses at the joint level to better understand the mechanism for decreased energetic efficiency in patients with MS.

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