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Comparison of Motor Control Deficits During Treadmill and Overground Walking Poststroke

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Abstract

Background. Force-sensing split-belt treadmills (TMs) provide an alternative to the conventional overground (OG) setting and allow new avenues for analyzing the biomechanics and motor control of walking. However, walking control may differ on a TM compared with walking OG. *Objective.* To compare spatiotemporal, kinematic, and EMG-based measures of motor control between TM and OG walking at self-selected and fastest comfortable speeds in persons with poststroke hemiparesis. *Methods.* Individuals with chronic hemiparesis (56) and similarly aged healthy individuals (17) walked over an instrumented walkway and on an instrumented split-belt TM; 16 channels of EMG recorded bilateral muscle activity, and a 12-camera motion capture system collected bilateral 3D kinematics. The authors applied a nonnegative matrix factorization (NNMF) algorithm to examine the underlying patterns of motor control. *Results.* Self-selected walking patterns differed on the TM versus OG in controls: speed decreased, stride length decreased, stance percentage increased, and double-support percentage increased. Poststroke, responses were similar, but cadence also decreased, and step length asymmetry increased. Kinematic patterns were similar except those associated with slower walking speeds. NNMF demonstrated similar EMG variance in the 2 environments. *Conclusion.* Persons, both healthy and poststroke, walk with different gait parameters on the TM. Although measures of motor control were mostly similar between the 2 environments, the TM induced step length asymmetry in 30% of participants (60% of whom took longer paretic steps). TM walking, therefore, is a valid method for detecting motor control deficits.

Keywords

walking, stroke, motor control, treadmill, overground

Introduction

A thorough understanding of the motor control and biomechanical deficits underlying walking dysfunction in persons with poststroke hemiparesis has been hampered by technological shortcomings in conventional gait laboratories. Specifically, it is difficult to collect enough trials of steady-state bilateral kinesiological data (eg, ground reaction forces) to robustly represent the disabled motor patterns. As a result, the bilateral walking pattern is more commonly interpreted by combining nonconsecutive unilateral single-step records assumed to represent the steady-state pattern.

Force-sensing treadmills (TMs) provide an alternative to the conventional overground (OG) setting and allow new avenues for analyzing the biomechanics and motor control of walking.[1-3] The most important advantage of an instrumented TM is that bilateral kinematics, kinetics, and EMG can be simultaneously and continuously measured for a large number of consecutive cycles to increase the signal-to-noise ratio and improve data quality. The availability of a large number of individual cycles also facilitates novel analysis techniques that use cycle-to-cycle variability to probe the underlying motor control structure.[4,5]

A potential drawback to the use of force-sensing TMs is the possibility that walking control differs on a TM compared with OG. Harris-Love et al[6] found that the TM induced a more consistent and temporally symmetric gait pattern in

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hemiparetic individuals walking at their self-selected OG speed, producing immediate changes in EMG symmetry.[7] Note, however, that participants held handrails in both studies, which fundamentally changes the task demands by constraining upper-body and trunk motion.[8] Bayat et al[9] tested individuals walking at self-selected speeds on a TM without handrails and found slower walking with shorter stride lengths and faster cadence on the TM than OG. Although these studies noted differences in spatiotemporal, kinematic, and EMG measures, they did not look for differences between measures of motor control in the 2 environments.

Although no measure is currently considered the standard for revealing a motor control deficit, several measures exist that can be used to test the hypothesis that TM walking is a valid method for assessing motor control deficits in persons with poststroke hemiparesis. Step length asymmetry,[10] measured by paretic step ratio (PSR; defined as the percentage of the stride length accounted for by the paretic step length) and highly correlated with paretic propulsion,[11] could serve as a spatiotemporal measure of motor control deficit. Midstance knee angle, which both Mulroy et al[12] and De Quervain et al[13] found to be a dominant kinematic variable for differentiating between locomotor control strategies, could serve as a kinematic measure of motor control deficit. The percentage of the cycle-by-cycle variance of lower-extremity EMG explained by locomotor control modules (coactive muscle groupings that are predictive of walking function poststroke) determined by nonnegative matrix factorization (NNMF)[5] can provide an EMG-based measure of motor control deficit.

The primary purpose of this study was to compare spatiotemporal, kinematic, and EMG-based measures of motor control deficit between TM and OG walking at self-selected and fastest comfortable speeds in persons with poststroke hemiparesis. Specifically, we hypothesized that self-selected TM walking will show differences compared with OG walking (most notably walking slower and having shorter stride lengths) but that indices of motor control deficits will remain consistent between the 2 walking modes. If our hypothesis is supported, this will establish TM walking as a valid method for assessing motor control deficits during walking.

Methods

Participants

A total of 56 individuals with chronic hemiparesis (61.0 ± 12.3 years, 5.1 ± 5.6 years poststroke, 20 women, 36 with left-side hemiparesis, self-selected walking speed ranging from 0.14 to 1.00 m/s, and Fugl-Meyer lower-extremity scores ranging from 8 to 34) and 17 similarly aged healthy individuals (age = 65.1 ± 10.4 years; 15 women) participated at the Malcom Randall VA Medical Center in Gainesville, FL. Inclusion criteria for persons poststroke were: hemiparesis secondary to a single onset unilateral stroke; ability to ambulate independently with or without an assistive device over 10 m on a level surface; ability to walk on a regular basis at home; absence of significant lower extremity joint pain and major sensory deficits; absence of significant lower limb contractures; and no significant cardiovascular or respiratory symptoms contraindicative to walking. Exclusion criteria were: any orthopedic or neurologic conditions in addition to stroke; significant musculoskeletal problems other than stroke that limit hip and knee extension or ankle plantar flexion to neutral; or inability to provide informed consent. All participants provided informed consent and the study was approved by the institutional review board and the University of Florida and the VA Subcommittee for Clinical Investigation.

Experimental Protocol

Participants walked on an instrumented split-belt TM (TECMACHINE, Andrezieux Boutheon, France) and through the same calibrated motion capture space OG. For TM walking, three 30-s trials were collected at self-selected TM speed after achieving steady-state and patient-stated comfort on the TM. For OG walking, 3 self-selected speed and 2 fastest comfortable speed trials were collected over a 16-foot instrumented walkway (GAITRite, Havertown, Pennsylvania). Speeds were independently chosen by the participants without external control exerted by the researchers. TM walking always preceded OG walking.

A safety harness system mounted to the ceiling above the TM protected participants in case of balance loss while walking on the TM. The harness off-loaded no body weight. A physical therapist or research assistant stood near the participants but did not provide assistance as they walked on the TM. The therapist also walked alongside the participants to provide safety during OG walking when necessary. Hemiparetic participants walked without using an assistive device or ankle-foot orthosis. Participants were provided with an Aircast (DJO, Vista, California) brace to provide inversion–eversion stability if necessary.

Data Collection

Data collection was done in a manner similar to that for other studies in our laboratory[14,15]. Markers were placed on participants in a modified Helen Hayes configuration, with rigid clusters of retroreflective markers placed on the feet, left and right shanks, left and right thighs, and pelvis to allow for 6 degrees of freedom of orientation measurement of each segment. Kinematic data were collected with a 12-camera Vicon (Vicon, Los Angeles, CA) system (100-Hz sampling rate).

Bipolar Ag–AgCl surface electrodes were used to record EMG from the tibialis anterior, soleus, medial gastrocnemius, vastus medialis, rectus femoris, medial hamstrings, lateral hamstrings, and gluteus medius of each leg using a telemetered EMG system (Konigsberg Instruments, Pasedana, California). Each skin site was shaved prior to electrode placement. All data were collected at 2000 Hz using Vicon software and saved to disk for analysis.

Kinematic and Spatiotemporal Data Processing

Raw kinematic data were low-pass filtered using a fourth-order, zero-lag Butterworth filter with a 10-Hz cutoff frequency. A custom MATLAB algorithm was written to identify gait cycle events from the video for both OG and TM walking. Step length was calculated as the distance between consecutive heel positions, and midstance knee angle was defined at the exact time midpoint of the single stance phase.

The average number of steps in the capture volume was 8.5, with an average of 4.9 used per trial. To be consistent, we calculated the spatiotemporals with the same algorithm in each mode as opposed to using a motion system–based measurement for the TM and using the output of the instrumented mat for OG walking. Note that more steps were measured during the OG walking by the GAITRite mat, even when not visible to the motion capture system. Quality control comparison of the instrumented mat (including all steps) and motion-based measurements of the same trials convinced us that we had characterized well the stepping pattern with the number of steps analyzed.

EMG Data Processing

EMG data were high-pass filtered (40 Hz) with a zero-lag, fourth-order Butterworth filter, debiased, rectified, and smoothed with a zero-lag, fourth-order, low-pass (4-Hz) Butterworth filter. To facilitate comparisons between participants and between TM and OG walking, the EMG from each muscle was normalized to its peak value during self-selected TM walking. We applied a NNMF algorithm to examine the underlying patterns of muscle activity in each condition.[16,17] The output of NNMF includes a matrix of muscle weightings, which represent coactive muscle groupings and a matrix of activation timing profiles that indicate when in the gait cycle the muscle groupings were active. NNMF decomposition yielded a small number of independent factors from the TM EMG data, which cumulatively accounted for at least 90% of the variability in each muscle and in each region of the gait cycle.[5] NNMF was then performed on the OG EMG data, but in this case, the muscle weightings identified during TM walking remained fixed, such that only the activation timing profiles could be altered to match the OG walking EMG. Similarly, NNMF with fixed muscle weightings was also applied to the third TM trial (which was not used in the determination of the muscle weightings) for the same number of steps analyzed during OG walking for that participant. Thus, we could test how much of the variability in the 2 data sets (with equal number of steps) was accounted for by the NNMF analysis.[5] A range of studies have found NNMF or similar decomposition techniques to be able to compare the motor control for different tasks and varied conditions within a single task, in a quantitative way.[18-20] Traditional EMG processing methods do not provide a similar method for quantitatively testing for differences in 8 channels of EMG data from multiple steps.

Data Analysis

Paired *t* tests were performed to test for differences between walking OG versus on the TM for all temporal, spatiotemporal, kinematic, and EMG measures. Although significance was assumed for $P < .05$, the actual P value is reported to allow for assessment of the strength of evidence. All calculations were performed with SPSS v17.0 (SPSS, Inc, Chicago, Illinois).

Results

Spatiotemporals

Controls walked differently at their self-selected speed on the TM in comparison with OG. They walked more slowly on the TM (TM, 1.06 vs OG, 1.24 m/s; $P = .0009$). Whereas cadence was unchanged (TM, 116.8 vs OG, 116.7 steps/min; $P = .99$), stride length decreased (TM, 1.04 vs OG, 1.30 m; $P < .0001$). Percentage of the gait cycle spent in stance increased (TM, 68.3 vs OG, 63.5 %; $P < .0001$), with the double-support phase also increased (TM, 36.6% vs OG, 27.1%; $P < .0001$; evenly split between the 2 double-support phases) more than the single-support phase decreasing (TM, 31.7 vs OG, 36.5 %; $P < .0001$). Asymmetry (PSR with the right leg assigned to be the "paretic" leg) was relatively minor during both TM and OG walking in controls (TM, 0.502 ± 0.015 vs OG, 0.501 ± 0.012). There was no difference between modes ($P = .75$).

Step length asymmetry in the hemiparetic participants was classified using the level of symmetry in the controls. Given the variability in PSR demonstrated by the controls (standard deviation [SD] = 0.0115), we defined symmetric stepping for hemiparetic participants as the range from 0.465 to 0.535 (approximately perfect symmetry of 0.5 ± 3 SDs). This range encompassed all OG PSR values and all but 1 of the TM PSR values, and it represents approximately 7% asymmetry.

Those with hemiparesis also walked more slowly on the TM (TM, 0.38 vs OG, 0.58 m/s; $P < .001$), with increased cadence (TM, 84.9 vs OG, 77.6 steps/min, $P = .04$) and decreased stride length (TM, 0.52 vs OG, 0.85 m; $P < .0001$). Similar to controls, there was an increase in percentage of gait cycle spent in stance in both legs (paretic: TM, 72.1% vs OG, 65.6% and $P < .0001$; nonparetic: TM, 81.7% vs OG, 75.8 % and $P < .0001$), with the double-support phase increasing (TM, 25.9% vs OG, 18.1% and $P < .0001$ for the first paretic double-support phase; TM, 28.2 vs OG, 23.8 and $P = .008$ for the second double-support phase or paretic preswing). Temporal asymmetry was unchanged as reflected by the difference in single-limb support between the legs (nonparetic leg single-limb support vs paretic leg single-limb support: TM, 9.6% vs OG, 10.1%; $P = .38$). Absolute spatial asymmetry ($|PSR - 0.50|$), which we considered so that increased low PSR asymmetry

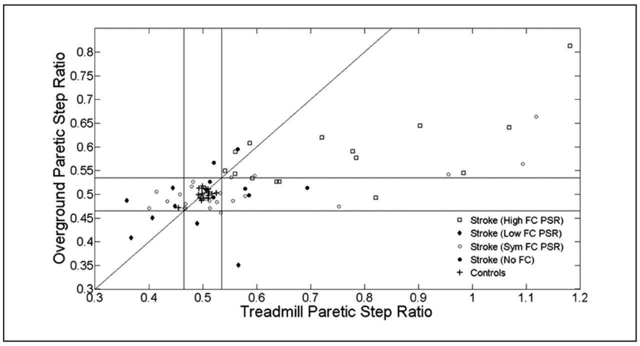

Figure 1. Comparison of self-selected speed paretic step ratio (PSR) for treadmill and overground walking: the vertical and horizontal bounded regions represent the defined boundary of the symmetric region (PSR between 0.465 and 0.535). The diagonal line represents equal values for treadmill and overground PSR. For overground PSR > 0.5, points to the right of the line represent more asymmetry on the treadmill, and for overground PSR < 0.5, points to the left of the line represent more asymmetry on the treadmill. The shape of the symbol represents the asymmetry group for that participant when walking overground at his or her fastest comfortable speed. Participants were nearly all either in the same PSR group in both conditions or became asymmetric during treadmill walking. Also note that when participants were symmetric overground but asymmetric on the treadmill (in horizontal shaded region), they often exhibited the same asymmetry during fastest comfortable walking overground.

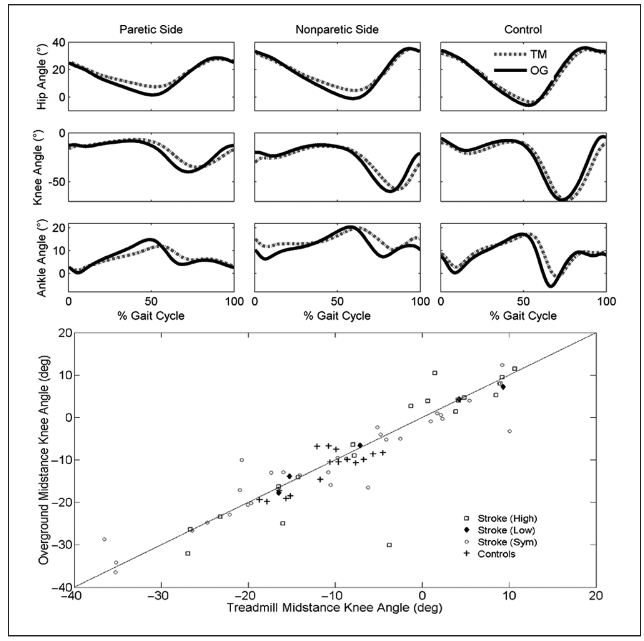

Figure 2. Comparison of lower-extremity joint kinematics for treadmill (TM) and overground (OG) walking: there were remarkable similarities between conditions in controls and in the paretic and nonparetic legs of hemiparetic individuals. There was reduced peak hip extension in TM walking (minimum blue value) consistent with a shorter stride length. Midstance knee angle was relatively unchanged (inset).

does not cancel out increased high PSR asymmetry, increased for the hemiparetic individuals as a group (TM, 0.159 vs OG, 0.053; $P < .0001$). When we used PSR to classify participants into groups with different motor control deficits (high PSR, low PSR, or symmetric), we found that the vast majority of participants (93%) either showed the same deficit or went from symmetric OG to an asymmetric deficit on the TM that was not revealed OG at a self-selected speed (Figure 1). Also, 62% of participants displayed the same category of asymmetry during TM walking as they did during OG walking (35 of 56 participants: 2 of 5 low-PSR participants, 15 of 32 symmetric participants, and 18 of 19 high-PSR participants). Slightly more than half (53%) of the participants who were symmetric OG became asymmetric on the TM (10 high PSR and 7 low PSR). It is interesting to note that 43% of participants who became asymmetric (of those for whom we have fastest comfortable OG PSR data) on the TM also showed the same asymmetry when walking OG at their fastest comfortable speed (6 of 14 participants: 3 of 7 participants who became low PSR, 3 of 7 participants who became high PSR; we did not have fastest comfortable speed PSR data for the other 3 participants); 5% of participants who were asymmetric OG became symmetric on the TM (1 high PSR and 2 low PSR; Figure 1). One participant who was asymmetric OG switched asymmetric groups on the TM (low PSR OG and high PSR on the TM).

Kinematics

The kinematics of walking on the TM were mostly similar to walking OG, with the exception of differences that were consistent with the changes of walking with a faster cadence and shorter step length (Figure 2). For both control and hemiparetic participants, there was a decrease in peak hip extension in late stance and an increase in peak knee flexion during swing and increased peak dorsiflexion in late stance. Our kinematic measure of motor control (midstance knee angle; 0° is full extension and negative values represent flexion) was unchanged between modes in controls (TM, −11.3° vs OG, −12.4°; $P = .15$) and the paretic legs of hemiparetic participants (TM, −8.2° vs TM, −8.9°; $P = .29$).

Electromyogram

The paretic leg EMG of walking on the TM was mostly similar to that of walking OG (Figure 3). Our EMG-based measure of motor control (variance accounted for [VAF] by the same NNMF-defined modules, see Figure 4, for example, control) exceeded our threshold of 90% in both hemiparetic participants and controls, respectively, during the TM (91.9% ± 4.1%, 93.5% ± 3.5%) and OG (95.2% ± 2.8%, 97.5% ± 1.0%) walking conditions at

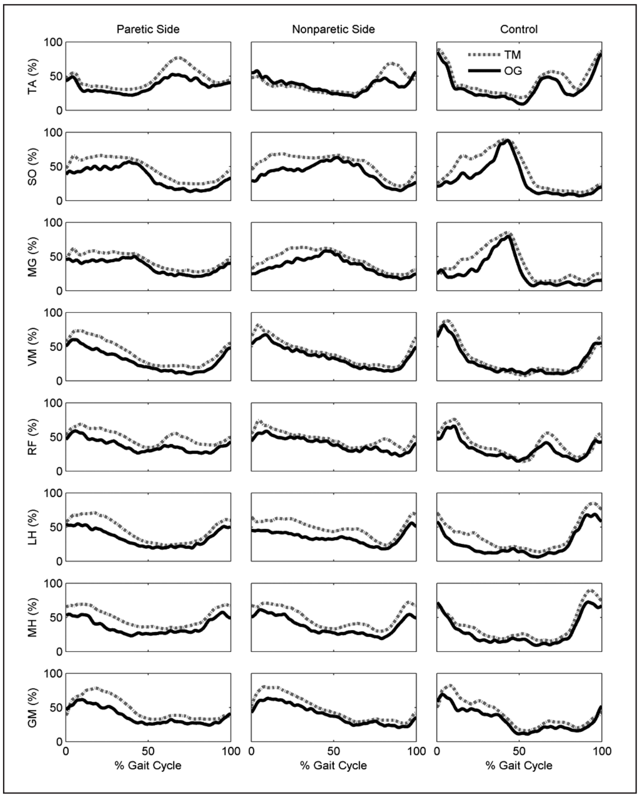

Figure 3. Comparison of lower extremity EMG for treadmill (TM) and overground (OG) walking: TM walking (blue lines) tended to elicit greater magnitude of muscle activity than did OG walking (red lines). In general, the shape of the muscle activation profiles tended to be similar between modes.

self-selected speed. Therefore, the same patterns of muscle activity observed during TM walking could also accurately describe the muscle activity during OG walking (see Figure 5 for OG and TM comparisons in those with hemiparesis).

Discussion

Both TM and OG walking were characterized by similar temporal asymmetry, midstance knee angle (kinematic measure of motor deficit),[12,13] and amount of variability accounted for by the same NNMF-derived modules (EMG measure of motor deficit).[5] The primary difference between TM and OG walking is that participants walked more slowly, with shorter strides, and with more time spent in double support. Consistent with slower walking and shorter strides, there were kinematic differences such as decreased peak hip extension, decreased knee flexion during midswing, and decreased peak ankle dorsiflexion. We also found strong evidence that TM walking increased step length asymmetry (spatiotemporal measure of motor deficit) poststroke. Notably, the increased step length asymmetry observed during TM walking was very similar to the increased step length asymmetry observed during fastest OG walking. This finding suggests that TM walking provided a challenge similar to walking as fast as possible and exacerbated the hemiparetic participants' existing motor control deficit. Although there are some differences in the movement patterns between TM and OG walking, the evidence presented here supports the assertion that TM walking at self-selected speed serves as a valid method for detecting and quantifying the motor control deficits present during OG walking in persons with poststroke hemiparesis.

Spatiotemporal Motor Control Deficits

Whereas previous literature has been inconsistent regarding whether stepping asymmetry differs between TM and OG walking, this study definitively establishes that there is no immediate increase in temporal or spatial symmetry when persons with hemiparesis walk naturally on the TM (ie, no external constraints such as holding a handrail) because step length asymmetry increased on the TM. Early studies comparing hemiparetic walking on the TM with OG walking found a difference between TM and OG walking,[6,7] but their results were likely influenced because participants were allowed to hold a handrail at the front of the TM.[9] The introduction of this kinematic constraint could account for much of their findings because holding a handrail limits the extent to which a person's body can travel backward on the TM belt. Bayat et al[9] did not find an immediate change in temporal asymmetry in their study of 10 hemiparetic individuals. Their results were very similar to ours, in that they found decreased self-selected speed with increased cadence and decreased stride length. However, they did not test step length asymmetry. Our study provides strong evidence that step length asymmetry[10,11] increases during TM walking.

Despite increases in step length asymmetry, which is our spatiotemporal measure of motor control deficit, we believe that self-selected TM walking can serve as a valid method for determining poststroke motor control deficits. Step length asymmetry revealed very few instances of hemiparetic individuals showing changes from one type of motor control deficit to another (ie, change from taking significantly longer steps with the paretic leg to longer steps with the nonparetic leg, or vice versa). Differentiating participants into high-PSR, symmetric, or low-PSR groups based on OG walking, we found that 91% either remained in the same asymmetry group or moved from the symmetric to an asymmetric group. Indeed, we found that only 1 participant changed from 1 asymmetric PSR group to the other (from low PSR OG to high PSR on the TM). Note that this participant showed an uncharacteristically high value of paretic propulsion (0.69) during TM walking (paretic propulsion measurements were unavailable during OG walking), which is more commonly associated with low PSR walking,[10] suggesting that their TM and OG walking mechanics were

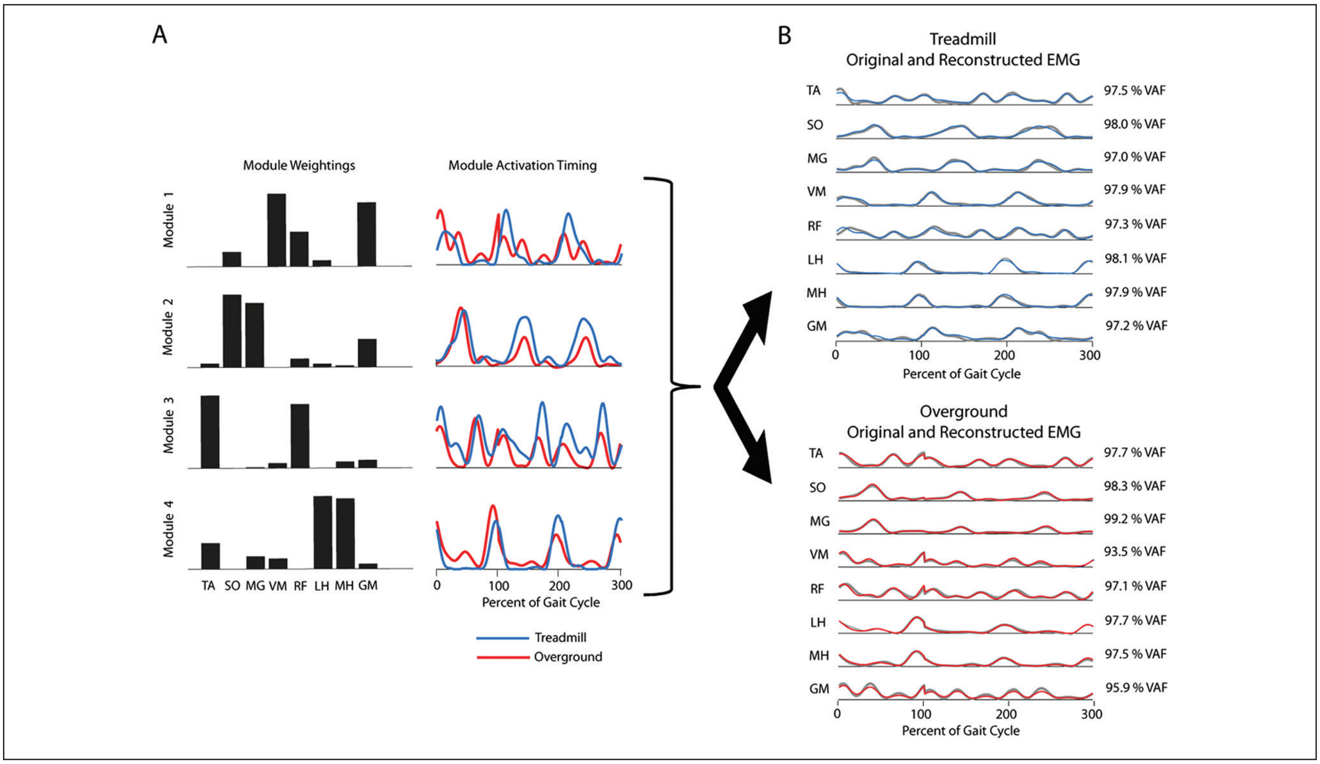

Figure 4. Reconstruction of EMGs by nonnegative matrix factorization (NNMF) for a representative control participant: 3 consecutive gait cycles are shown from walking on a treadmill (blue) and overground (red). A. Module weightings determined from NNMF of treadmill EMG (panel A, left) were used to optimally reconstruct the EMG from separate sets of treadmill and overground walking. The resultant timing of each module is shown on the right-hand side of panel A. B. Original EMG, reconstructed EMG (in gray), and the percentage of variability accounted for (VAF) by the reconstructed EMG for each muscle's EMG during treadmill (panel B, top) and overground walking (panel B, bottom). Abbreviations: TA, tibialis anterior; SO, soleus; MG, medial gastrocnemius; VM, vastus medialis; RF, rectus femoris; LH, lateral hamstrings; MH, medial hamstrings; GM, gluteus medius.

likely more similar than might be expected by the change from low to high PSR. A particularly intriguing finding was that 53% of the participants who walked symmetrically OG became asymmetric on the TM, and nearly half of those participants were also found to be similarly asymmetric when walking at their fastest comfortable speed OG. Therefore, the TM environment appeared to provide a challenge beyond steady-state walking and revealed a motor control deficit that was not evident during self-selected OG walking.

Kinematic Motor Control Deficits

Midstance knee angle was unchanged between the 2 modes of walking in both controls and persons with hemiparesis. Midstance knee angle was analyzed because it has been used previously as a kinematic measure to classify participants into groups with similar gait patterns.[12,13] Midstance knee angle is a particularly good measure because it is unlikely to be speed dependent.[12,13] The fact that midstance knee angle did not change supports the hypothesis that the primary motor control deficit is similar between modes.

EMG Motor Control Deficits

Using NNMF, we have demonstrated that the same small set of factors that account for muscle activity during TM walking also account for muscle activity during OG walking. This finding supports the notion that both conditions involve the same fundamental motor control strategy. Previous studies have used similar factorization methods to determine if the same patterns of motor output are used to produce different tasks[21,22] and to induce adaptations to the locomotor pattern.[18] The similar amounts of VAF imply no substantial changes in the relative magnitude and timing of muscle activity between TM and OG walking. It is not surprising that the TM condition had higher VAF because the factorization was performed on TM walking data (although the steps analyzed in the comparison were from a separate trial that was not used in the factorization). Nevertheless, because the VAF was greater than 90% in both modes using the same modules, we conclude that the factorization described both modes of walking very well and that there was no evidence of a difference in motor control deficits between modes.

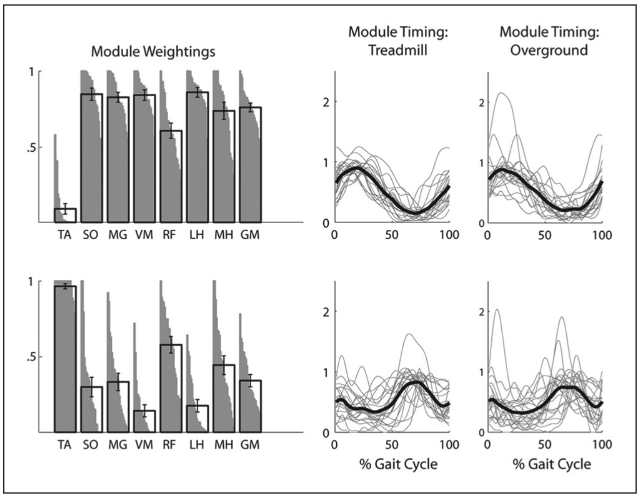

Figure 5. Comparison of module muscle weightings and activation timing profiles for the paretic leg of persons poststroke that exhibited 2 modules for treadmill and overground walking: the muscle weightings for the 2 modules were calculated from 2 trials of treadmill walking for each participant ("Module Weighting" panels). Nonnegative matrix factorization (NNMF) was then performed on the overground EMG data, but in this case, the muscle weightings identified during treadmill walking remained fixed such that only the module timing profiles could be altered to match the overground walking EMG ("Module Timing: Overground" panels). Similarly, NNMF with the same fixed muscle weightings was also applied to the third treadmill trial (note that this trial was not used in the determination of the muscle weightings) for the same number of steps analyzed during overground walking for that participant ("Module Timing: Treadmill" panels). The group averages of the 2 different module timing curves were remarkably similar for treadmill and overground walking. Abbreviations: TA, tibialis anterior; SO, soleus; MG, medial gastrocnemius; VM, vastus medialis; RF, rectus femoris; LH, lateral hamstrings; MH, medial hamstrings; GM, gluteus medius.

Similarities Between TM and OG Walking

In addition to theoretical analyses that demonstrate that the physics of OG walking and TM walking are the same as long as the belt moves at constant velocity,[23] multiple studies with healthy participants have demonstrated that individuals can walk on the TM with essentially the same mechanics that they use to walk OG. We have shown that there was no difference in propulsion generated by participants when they walked on a TM when speed and stride length are carefully controlled to be the same as OG.[24] Riley et al[25] concluded that while there were differences in the many measures of peak moments and powers during TM walking, they should not be considered meaningful because they were still within the range of normal variability. Stoquart et al[26] investigated TM walking at multiple speeds and concluded that they could rule out "that biomechanics of TM and OG walking could be different." Tesio and Rota[3] found that dynamic, kinematic, and EMG measures were similar between walking modes in a study in which they used independent samples for each mode. They concluded that gait analysis of TM walking was a potentially valuable method of gait analysis in a clinical setting. Lee and Hidler[27] concluded that even though there were differences in EMG, joint moments, and joint powers between the 2 walking modalities, the overall patterns of these measures were quite similar. Parvataneni et al[28] investigated the 2 modes at matched speed in an elderly population and found that the modes were mostly similar in kinematics and dynamics, although the metabolic cost was 23% higher during TM walking. Bollens et al[29] found that the long-range autocorrelations present among stride duration variability were also similar between the 2 modes of walking. A synthesis of the literature leads to the conclusion that TM walking at a matched speed to OG walking results in increased cadence (eg, Watt et al[30]) but demonstrates mostly similar patterns of kinematics, dynamics, and EMG. However, when allowed to self-select TM speed, participants usually walk more slowly with faster cadence. Thus, we conclude that TM and OG walking mechanics are similar, especially at matched speeds without the use of handrails or other kinematic constraints.

Conclusion

In conclusion, in the largest study to date (n = 56 hemiparetic participants), we conclusively found that hemiparetic participants do not demonstrate immediate improvements in symmetry (neither temporal nor spatial) when they are provided with no support (either a hand hold or body weight support). Instead, our results suggest that TM walking provided a challenge similar to walking as fast as possible and exacerbated the hemiparetic participant's existing motor control deficit, such that step length asymmetry increased. The changes that were observed between the 2 modes of walking, primarily in spatiotemporal measures (and the associated kinematics that accompany slower walking speed and shorter strides), did not influence our kinematic and EMG measures of motor control deficits, suggesting that walking on the TM reveals information relevant for understanding OG walking. Thus, we believe that walking on an instrumented TM at a self-selected speed can be a valid method for determining motor control deficits and walking impairments exhibited after stroke. Furthermore, the exceptional benefits provided by testing on an instrumented TM, most notably the ability to measure bilateral kinematics, kinetics, and EMG continuously for a large number of consecutive cycles in order to precisely determine the steady-state walking pattern and its associated step-to-step variability, allow us to substantially increase our understanding of hemiparetic walking. Investigators may also wish to record the PSR of self-selected and fastest comfortable speed during OG walking as it may provide valuable additional measures for patient classification.

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Declaration of Conflicting Interests

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