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Electromyographic and kinematic nondisabled gait differences at extremely slow overground and treadmill walking speeds

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Abstract—This study compared the kinematic and electromyographic (EMG) gait patterns of able-bodied adults at natural speed in contrast to extremely slow overground and treadmill walking speeds. Kinematic and EMG data were collected at three speeds (self-selected, 0.30 m/s, and 0.20 m/s). Eighteen subjects were evaluated for trunk and lower-limb motion and EMG of five lower-limb muscles. Significant reductions were found in segmental motion between natural speed and both slower gait speeds, accompanied by an expected reduction in cadence and stride. EMG patterns at slower speeds showed changes in timing and reduced magnitudes. Phasic timing of the proximal muscles showed the most changes with predominant coactivation, whereas the distal muscles remained consistent with the pattern at natural self-selected speed. Overground versus treadmill gait patterns revealed minimal differences. Consideration of the effects of slower walking speed may help clinicians create interventions to target primary gait deficits on overground or treadmill walking.

Key words: electromyography, gait analysis, gait training, kinematics, motion, nondisabled, normative data, overground/treadmill gait, rehabilitation, slow speed comparisons.

INTRODUCTION

Recovery of optimum walking function is a major goal of rehabilitation. A variety of human neurological deficits may cause a substantial decrease in walking speed. Observational and quantitative gait analysis is often limited by the lack of normal data at the extremely slow walking speeds seen particularly in patients with upper-motor-neuron deficits. Furthermore, gait analysis and training may involve both overground and treadmill conditions. This study compared kinematic and electromyographic (EMG) data of able-bodied subjects walking at self-selected and extremely slow speeds on a motorized treadmill and overground. These data will help to identify and isolate the effect of walking speed of rehabilitation patients to provide more insight for gait analysis and training in the rehabilitation population.

Abbreviations: EMG = electromyographic, MG = medial gastrocnemius, MH = medial hamstring, RF = rectus femoris, ROM = range of motion, SCI = spinal cord injury, SD = standard deviation, TA = tibialis anterior, VL = vastus lateralis.

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Normal walking speed in the able-bodied adult population is estimated to be 1.40 m/s (± standard deviation [SD] of 0.20) [1–2], while walking speeds of persons with hemiparesis following a stroke have been reported at approximately 0.40 m/s [3–4]. Olney and Richards considered 0.64 m/s to be fast for this population, 0.41 m/s to be average, and 0.25 m/s to be slow [5]. Similarly, walking speed for ambulatory spinal cord injury (SCI) patients may vary from the more severely injured, at 0.20 m/s, to moderately disabled at 0.60 m/s [6]. Krawetz and Nance [7] and Melis et al. [8] reported an average speed of 0.80 m/s in higher-functioning SCI patients.

Biomechanical research of slower walking speeds is often at faster speeds than the patients with neurological deficits. Normative studies have investigated slower walking speeds ranging from 0.60 m/s to 0.80 m/s [9–14]. The slowest speed measured by Murray et al. was 0.83 m/s [10], while Yang and Winter chose 0.75 m/s to represent slow walking [11]. Nilsson et al. [12] and Shiavi et al. [13] reported on a slow walking speed of 0.60 m/s. Common to all these reports were the observations that EMG and movement patterns decreased in amplitude with a decrease in walking speed [11–14]. The phasic patterns of EMG were less affected by the slower walking speeds. However, two studies reported that at slower speeds, the EMG phasic patterns were more variable, particularly at the hip and knee, compared with a comfortable, self-selected walking speed [9,11]. The EMG phasic timing of the ankle muscles remained more consistent at the slower walking speeds [11].

Lower-limb muscles may adapt to changes in walking speed by altering both activation timing and the EMG magnitude. These adaptations are muscle-specific [11]. However, little is known about gait pattern changes when nondisabled subjects adopt extremely slow walking speeds below 0.40 m/s. Although past research has shown that moderately slower normative gait (approximately 0.60 m/s) reveals lower amplitude in kinematic and EMG signals with slight disruption in muscle activation, no data exist on normative gait patterns at walking speeds of 0.30 m/s or 0.20 m/s. Authors have suggested that ambulating at speeds less than 0.30 m/s may be more shuffling than walking [13]. Neurological rehabilitation patients often demonstrate these extremely slow walking speeds.

In addition to slow walking speeds, rehabilitation patients are often evaluated and trained on a motorized treadmill. The differences between overground and treadmill gait at extremely slow walking speeds need to be identified to provide more complete information for clinical decision making. The kinematic and EMG gait differences between overground and treadmill walking at natural speeds have been studied and can be summarized. In contrast to overground walking, the gait pattern on a treadmill has shown (1) increased cadence and decreased stride length [15–16]; (2) full foot, in contrast to heel-strike, at initial contact [17]; (3) increased hip and knee flexion in swing with decreased knee extension in stance [15–16]; (4) forward trunk lean [16,18]; (5) relatively increased EMG amplitude of some lower-limb muscles, but the same phasic patterns compared with overground [16,19].

In summary, our primary goal of this study was to identify the effects of reduced walking speeds (0.30 m/s and 0.20 m/s) on overground and treadmill walking patterns in a nondisabled adult population. This information would help isolate the effect of slower speed of walking from the primary physical impairments that disrupt gait in the neurological patient.

METHODS

Subjects

Following approval by the Rehabilitation Centres Research Ethics Board, written consent was obtained from 18 able-bodied volunteers (13 females, 5 males). The subjects were in good health, with no conditions that affected their gait pattern or tolerance to the evaluation protocol. The age of subjects ranged from 23 to 58 years and weight ranged from 52 to 87 kg. While the subjects are few, previous work has used small groups for within-subject analysis and found appropriate significance [9–11]. A limiting factor for our pool of able-bodied volunteers was resource-related.

Evaluation Protocol

All data were collected in the Gait and Motion Analysis Laboratory at the Rehabilitation Centre (Ottawa, Canada). Subjects walked overground on a 10 m walkway and on a motorized treadmill (Biodex Medical Systems, Inc., Shirley, NY). All subjects were tested in one session. Data from three trials for each speed were collected for all subjects: at a self-selected (natural) speed and at speeds of 0.30 m/s and 0.20 m/s (for a total of nine trials). To achieve these slower speeds, subjects walked between two light beams while a MacIntosh computer provided on-screen feedback of their walking speeds. We used no other instructions or external pacers to avoid artificially setting the participant's cadence or stride length. Data were collected when each subject had reached a steady state of being able to replicate the walking speed consistently (allowing a window of 0.02 m/s) over two trials. All EMG data were obtained from the right side of each subject. Bilateral footswitches were used to collect stride time and events. This protocol was repeated for the treadmill condition. The treadmill was set at each individual's mean natural, overground speed, as well as at 0.30 m/s and 0.20 m/s.

Kinematics

The Ariel Performance Analysis System (APAS) (Ariel Dynamics Inc., Trabuco Canyon, CA) calculated angles at the ankle, knee, hip, and trunk segments (60 Hz). Subjects wore dark shorts, shirt, and shoes. Reflective markers were placed on the acromion, greater trochanter, lateral tibial plateau, lateral malleolus, posterior aspect of the calcaneus, fifth metatarsal, and toe of the right side of the body. Two-dimensional motion was obtained by one Panasonic video camcorder located perpendicular to the line of progression as subjects walked along the 10 m walkway and on the treadmill.

Electromyography and Footswitches

EMG data were collected from five right lower-limb muscles: rectus femoris, vastus lateralis (VL), medial hamstrings, tibialis anterior, and medial gastrocnemius. We cleaned and prepared skin sites, overlying each muscle belly, to decrease resistance [20]. Preamplified electrodes (DelSys, Inc., Boston, MA) were placed at standardized locations for optimal EMG recording of muscle activation [21]. Amplified EMG signals were collected by the APAS software. Four footswitches, placed under the sole and heel of each shoe, provided information on foot contact and toe-off during the gait cycle.

Data Analysis

EMG data were filtered (10 Hz) and full-wave rectified into a linear envelope with the use of BIOPROC software (University of Ottawa, Ontario, Candada). Data were normalized to a percentage of a stride. EMG amplitudes were normalized to 100 percent of the mean value of the same muscle at natural overground speed [22]. This normalization enabled comparisons between the three different walking speeds on overground and treadmill. EMG data from four strides within each trial (three trials for a maximum of 12 strides) were averaged for each speed.

The mean range of motion (ROM) of the ankle, knee, hip, and trunk segments was averaged over three strides and SDs were obtained. Data were normalized to 100 percent of one stride for overground and treadmill conditions. Averages and SDs for walking speed (meters per second), cadence (steps per minute), stride length (meters), and stance times (expressed as a percentage of the stride) were calculated for the 18 subjects.

Significant differences between overground and treadmill gait parameters were tested with two-way paired *t*-tests, and significant differences between the three walking speeds (natural and two extremely slow walking speeds) were tested with a one-way analysis of variance (Tukey's post hoc). The a priori significance level was set at $p < 0.05$. The quantified EMG amplitudes were not tested for significance because of the variability of both the intersubject data and different methods of EMG processing and normalization in the literature.

RESULTS

Data on 18 subjects were complete for the temporal-spatial and EMG parameters. Technical and data collection difficulties limited the number of subjects with full data to10 subjects for the motion analysis.

Temporal-Spatial Parameters

Table 1 illustrates the temporal-spatial results of overground and treadmill walking at natural and extremely slow walking speeds (0.30 and 0.20 m/s). Variability of the mean cadence, stride, and stance periods were similar between overground and treadmill at the three walking speeds. However, the slowest speed (0.20 m/s) showed the greatest variability on overground and treadmill (**Table 1**). A statistically significant difference existed between overground and treadmill in cadence and stride length at natural speed only (cadence, $p = 0.004$; stride length, $p = 0.002$). At the two slower speeds, overground and treadmill did not significantly differ. However, all temporal-spatial parameters at both slower speeds were significantly different from these variables at natural speed ($p < 0.05$). No significant difference in temporal-spatial parameters existed between the two slower speeds.

Table 1.
Mean temporal-spatial gait parameters (n = 18) ± standard deviation on overground and treadmill at natural and slow walking speeds. Two-way paired t-tests compared overground and treadmill values and analysis of variance (ANOVA) (Tukey's post hoc) compared values across walking speeds.

Parameter	Overground	Treadmill	p-Value, Paired t-Test
Cadence (steps/min)			
Natural	112.3 ± 9.3	117.8 ± 7.2	0.004*
0.30 m/s	48.9 ± 8.7†	52.5 ± 9.4‡	0.14
0.20 m/s	41.3 ± 11.8†	41 ± 10‡	0.89
Stride Length (m)			
Natural	1.55 ± 0.13	1.47 ± 0.08	0.002*
0.30 m/s	0.74 ± 0.13†	0.71 ± 0.12†	0.26
0.20 m/s	0.65 ± 0.16†	0.63 ± 0.23†	0.76
Stance (%)			
Natural	62.1 ± 2.5	63.1 ± 2.4	0.13
0.30 m/s	72.7 ± 3.9†	71.4 ± 3.2†	0.19
0.20 m/s	74.8 ± 4.2†	73.3 ± 4.8†	0.23

*Significant difference between overground and treadmill values (two-way paired t-test).
†ANOVA Tukey's post hoc $p < 0.05$ from natural speed.
‡ANOVA Tukey's post hoc $p < 0.01$ from natural speed.

Table 2.
Mean range of motion (ROM) in degrees (n = 10) ± standard deviation of ankle, knee, hip, and trunk segments, overground and treadmill at natural and slow walking speeds. Two-way paired t-test compared overground and treadmill values and analysis of variance (ANOVA) (Tukey's post hoc) compared values across walking speeds.

Segments (ROM)	Overground	Treadmill	p-Value, Paired t-Test
Ankle (°)			
Natural	28.3 ± 5.0	30.9 ± 5.7	0.28
0.30 m/s	23.4 ± 7.8	20.6 ± 6.6*	0.10
0.20 m/s	23.9 ± 10.4	22.9 ± 8.5†	0.64
Knee (°)			
Natural	64 ± 4.8	61 ± 10	0.34
0.30 m/s	48 ± 3.9*	46.1 ± 4.3*	0.25
0.20 m/s	46.4 ± 6.7*	45 ± 5.9*	0.63
Hip (°)			
Natural	38.2 ± 6.4	38.3 ± 6.2	0.98
0.30 m/s	27.7 ± 6*	29.6 ± 4.5*	0.20
0.20 m/s	26.9 ± 6.1*	28.1 ± 2.7*	0.53
Trunk (°)			
Natural	10.9 ± 2.4	10.9 ± 3.1	0.98
0.30 m/s	4.4 ± 1.1*	6.3 ± 1.2*	0.001‡
0.20 m/s	4.3 ± 1.1*	5.4 ± 1.4*	0.09

*ANOVA Tukey's post hoc $p < 0.01$ from natural speed.
†ANOVA Tukey's post hoc $p < 0.05$ from natural speed.
‡Significant difference between overground and treadmill values (two-way paired t-test).

Segmental Motion

Table 2 illustrates the difference in variability and average total ROM at natural and the two slower walking speeds on overground and treadmill. **Figure 1** provides timing-specific information showing the mean kinematic patterns at the three walking speeds. A phase shift of the toe-off event is related to a difference in stance and swing durations at the slower walking speeds. Each segment is described in the following in terms of the timing and amplitude of the movement patterns in the stance and swing phases.

Ankle

Overground, at natural speed, ankle plantar flexion was observed after initial contact followed by dorsiflexion in midstance. However, a slight loss of plantar flexion occurred following initial contact at both slower speeds. In late stance, mean plantar flexion, at natural speed, was 9°, whereas at the two slower speeds, plantar flexion beyond neutral was absent. This change decreased the total ROM of the ankle from 28° at natural speed to 23° and 24° at 0.30 m/s and 0.20 m/s, respectively. The decrease at the ankle ROM was not significantly different. Ankle dorsiflexion in the swing phase was minimally decreased at the slower walking speeds.

On the treadmill, the ankle at all three speeds lost the expected plantar flexion posture following initial contact, indicating a more foot-flat contact in contrast to heel-strike. At natural speed, dorsiflexion in midstance was reduced compared with the overground condition. In late stance, the magnitude of ankle plantar flexion was slightly greater than overground and was substantially reduced at the slower speeds. The reduced ankle ROM at both slower walking speeds was significantly different from the ROM at natural speed ($p < 0.05$). Ankle dorsiflexion in swing appeared to be slightly diminished compared with the overground condition at natural speed. At both slower speeds, the ankle patterns in swing remained at 5° or near neutral. No significant differences existed between the overground and treadmill results of ankle ROM. However, despite the overall ROM, the motion

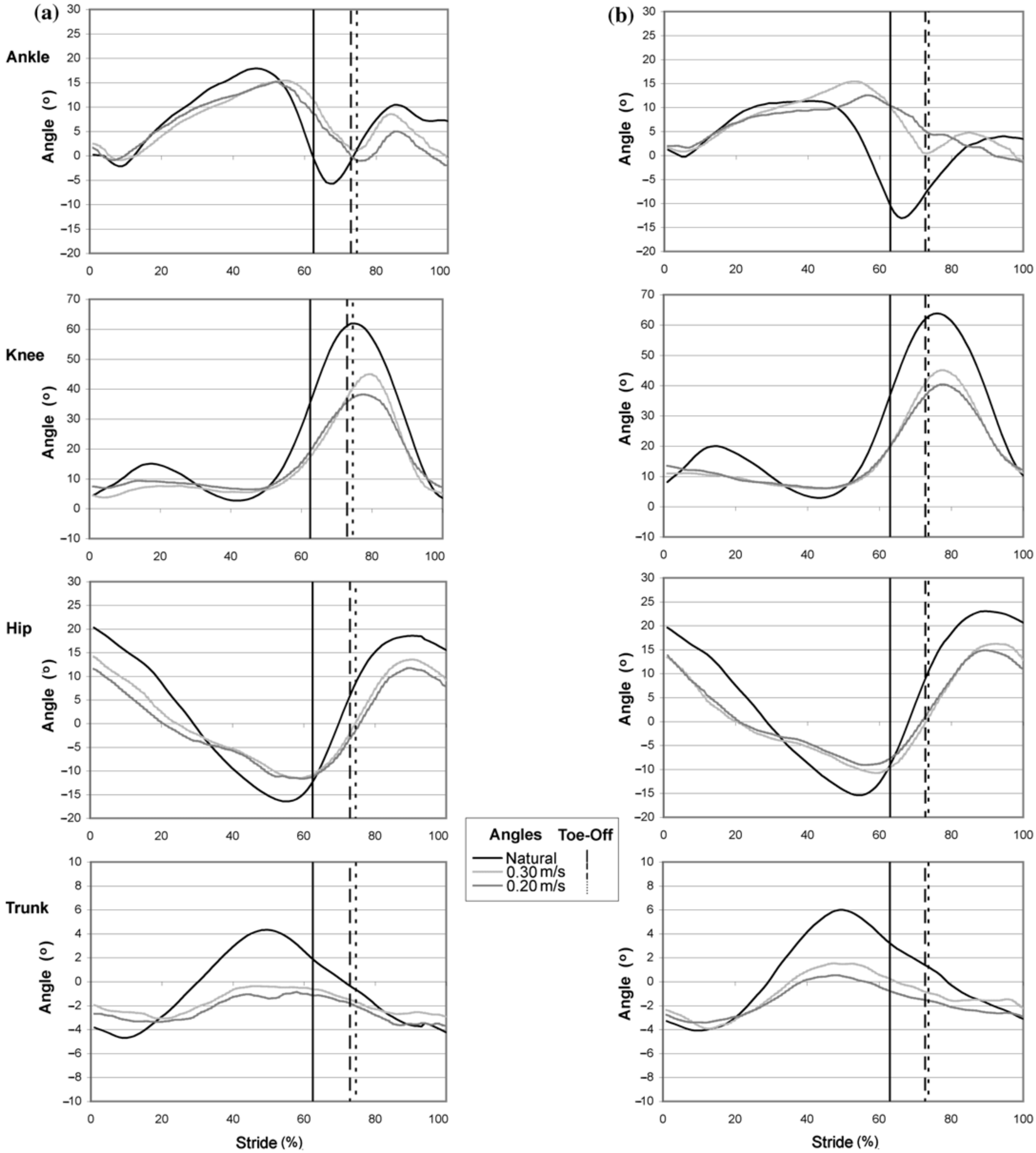

Figure 1.
Mean ankle, knee, hip, and trunk segmental motion (n = 10) normalized to 100% of one stride at three walking speeds: **(a)** overground and **(b)** treadmill. Vertical lines indicate toe-off event.

occurred in a different portion of the range, since dorsiflexion was decreased and plantar flexion increased for the treadmill walking condition.

Knee

Overground, at natural speed, knee flexion in early stance showed a typical 10° flexion/extension pattern during loading. This pattern was absent at the two slower walking speeds. At natural speed, mean knee flexion reached a peak of 61° in swing, while at the two slower speeds, a decrease of 16° and 23° occurred in mean peak amplitudes at 0.30 m/s and 0.20 m/s, respectively. These differences in knee ROM at both slower speeds compared with natural speed were statistically significant ($p < 0.01$).

On the treadmill, at natural speed, the knee posture at initial contact was slightly more flexed than on overground, but the typical knee flexion/extension on loading was maintained. At slower speeds, the knee pattern was similar to the overground condition, losing the flexion/extension loading response. Maximum knee flexion in swing at all three speeds on the treadmill was similar in timing and amplitude to the overground condition and showed a mean reduction of 19° and 24° at the two slower walking speeds. The decreased knee ROM was significantly different from the ROM at natural speed ($p < 0.01$) (**Table 2**). No significant differences existed between overground and treadmill knee ROM at the three walking speeds.

Hip

Overground, at initial contact, mean hip flexion at natural speed was approximately 20° but decreased by 5° to 8° at the slower speeds. In late stance, maximal hip extension at the natural speed was 17° beyond neutral but decreased by an average of 5° at the two slower speeds. During mid-swing, mean peak hip flexion attained 19° but was reduced by 6° at the two slower speeds. These reductions in hip ROM at both slower speeds were significantly different from the ROM at natural speed ($p < 0.01$).

On the treadmill, at natural speed, mean hip extension in late stance was similar in magnitude to the overground pattern, while mean hip flexion in swing was 6° greater than on overground. At the slower speeds, a reduction in both hip extension and flexion occurred. The reduced ROM at the slower walking speeds was significantly different from the ROM at the natural speed ($p < 0.01$). The overground and treadmill comparisons did not reveal significant differences across the three walking speeds.

Trunk

During overground walking, the mean trunk posture was slightly extended by 4° at initial contact and early stance with an average of 4° of trunk flexion during late stance. During swing, the trunk returned to the extended posture for initial contact. At the slower speeds, the trunk segment remained relatively extended during the entire gait cycle. The reductions in trunk ROM at both slower speeds were significantly different from the natural speed pattern ($p < 0.01$).

On the treadmill, at natural speed, the trunk segment showed the same extended posture at initial contact as observed in the overground condition. However, the trunk posture in late stance was slightly more flexed on the treadmill than on overground. At the slower walking speeds, this posture became more extended, as seen at the slower overground walking and was also significantly different from the pattern at natural speed ($p < 0.01$). The flexed trunk posture on the treadmill at 0.30 m/s was significantly different from the trunk posture at the same speed on overground ($p = 0.001$) (**Table 2**). However, the 2° difference was not beyond the margin of error.

Summary

The angular patterns were not different between the overground and treadmill conditions at natural speed. Minor trends in differences were observed on the treadmill:

- Slight decrease in ankle dorsiflexion in swing and a more foot-flat posture at initial contact.
- More knee flexion in stance.
- Slightly increased hip flexion in swing.
- Slightly increased trunk forward lean.

At slower speeds, both conditions showed statistically significant reductions in motion at the ankle, knee, hip, and trunk segments with intact timing of motion relative to the gait cycle.

Electromyography

Ensemble averaged EMG linear envelopes were calculated for each muscle of each subject and normalized to a percentage of a stride. The mean voltage of the normal speed linear envelope for each muscle of each subject was calculated. This mean voltage was then used to normalize the voltages for the subject's three walking speeds. These are displayed in **Figure 2** for overground and treadmill conditions. **Table 3** gives the mean peak EMG values expressed as a percentage of the mean EMG value of each muscle at natural overground speed.

Tibialis Anterior

Overground, at natural speed, the tibialis anterior (TA) muscle was typically biphasic, with the larger peak at initial contact corresponding to controlled plantar flexion at the ankle. The second peak followed toe-off and into early swing during foot clearance. At the two slower speeds, the EMG burst at initial contact and at toe-off was substantially lower than at natural speed, but the expected phasic pattern was still evident.

On the treadmill, at natural speed, a similar TA biphasic pattern emerged. The first peak at initial contact was relatively greater than on overground. The second peak was also similar to the overground pattern. At the slower speeds, the peaks similarly decreased at initial contact and at toe-off, but the main phasic pattern was maintained.

Medial Gastrocnemius

Overground, at natural speed, the medial gastrocnemius (MG) muscle was typically monophasic, and peak activity was approximately at 40 percent of the stride. A relatively quiet period was maintained in swing. At the two slower speeds, the EMG pattern showed a similar monophasic peak action during stance, but the amplitude was substantially reduced from natural speed.

On the treadmill, the EMG pattern at natural speed was similar to the overground phasing, with a relative increase in peak amplitude. At the slower speeds, the relative amplitudes of the EMG pattern were substantially reduced from the pattern at natural speed.

Rectus Femoris

Overground, at natural speed, the first and highest peak of the rectus femoris (RF) muscle occurred during early stance, while the second smaller peak was evident in midstance. The third RF peak in stance occurred at the transition from stance to the swing phase. In late swing, the RF was active in preparation for initial contact. At the two slower speeds, the three discrete peaks in stance were not obvious and were replaced by a lower-level EMG burst with a peak at approximately 30 percent of the stride during loading. In late swing, a rise in activity was evident prior to initial contact, but of lower amplitude than at natural speed.

On the treadmill, at natural speed, the three distinct RF peaks were similar to overground, with a relative decrease in peak amplitudes at weight acceptance and midstance compared with overground. At the slower speeds, the RF pattern showed relatively lower and less phasic EMG patterns compared with the overground pattern at the same slower speeds.

Vastus Lateralis

Overground, the mean EMG peak of VL activity at natural speed was in early stance. A second peak occurred in late swing in preparation for initial contact. At the slower speeds, the first peak in early stance was substantially reduced, with a relatively lower EMG phasic action during loading as was seen in the RF pattern. The phasing in late swing was still evident but with reduced amplitude compared with the natural speed.

On the treadmill, the EMG of the VL showed a phasic burst in early stance and late swing. However, the first peak was relatively reduced in amplitude compared with overground at natural speed. At slower speeds, the pattern was intact and similar to overground, with a substantial decrease in relative amplitude compared with natural walking speed.

Medial Hamstrings

On overground, at natural speed, the EMG of the medial hamstrings MHs muscle displayed a peak activity in mid to late swing and into early stance. At the slower walking speeds, the discrete phasic burst of EMG was substantially reduced both in late swing and at initial contact. The MH peak in early stance was replaced by a low-level burst in early to midstance during loading. An additional low-level EMG burst appeared at the transition from stance to swing, which was not observed at natural speed.

On the treadmill, the EMG pattern of MH at natural speed was similar in timing in mid to late swing when compared with the overground pattern but with relatively higher amplitude. The EMG amplitude was less in early stance than on overground. The patterns at slower speeds were similar to the overground patterns, with substantially reduced EMG amplitudes in late swing and a low-level EMG phasic burst at the stance-to-swing transition. However, the phasic pattern in early stance was similar in amplitude to overground natural speed.

Summary

The overground and treadmill EMG patterns at natural speed were similar except for the following differences related to the relative EMG amplitude changes on the treadmill:

- TA: slight increase in EMG at initial contact.
- MG: slight increase in relative peak amplitude.
- RF: decreased EMG in early to midstance.
- VL: decreased EMG in early to midstance.
- MH: increased peak in late swing and decreased peak in early stance.

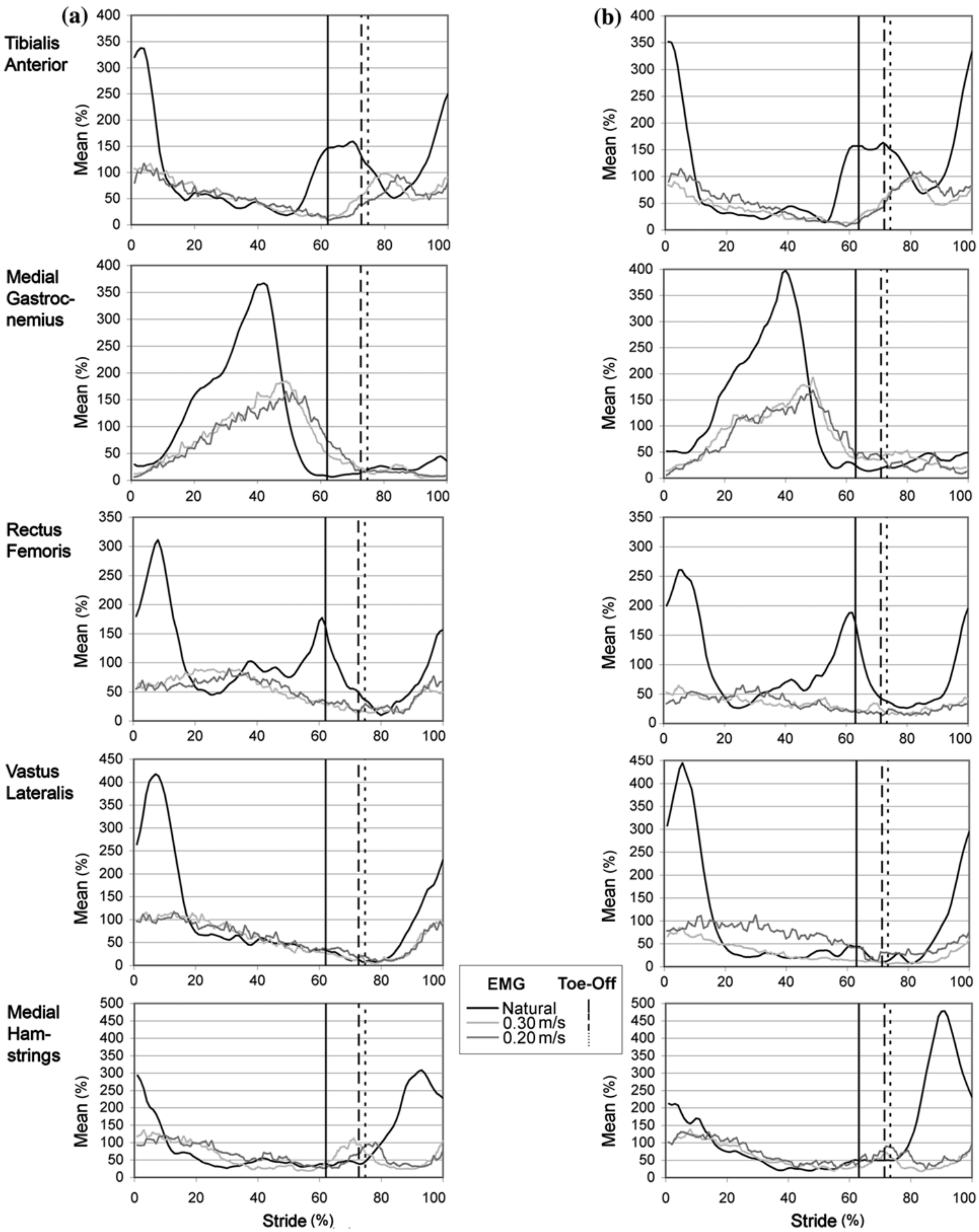

Figure 2.
Ensemble average electromyographic (EMG) patterns (n = 18) normalized to 100% of one stride at three walking speeds: **(a)** overground and **(b)** treadmill. Vertical lines indicate toe-off event.

Table 3.
Mean electromyographic (EMG) peak values (n = 18) expressed as a percent of mean EMG value (natural speed, overground) ± standard deviation. Values represent overground and treadmill conditions at natural and slow walking speeds.

Parameter	Tibialis Anterior	Medial Gastrocnemius	Rectus Femoris	Vastus Lateralis	Medial Hamstrings
Overground					
Natural	341 ± 119	367 ± 112	319 ± 151	417 ± 117	313 ± 83
0.30 m/s	115 ± 75	194 ± 160	87 ± 61	122 ± 110	105 ± 107
0.20 m/s	121 ± 93	185 ± 161	81 ± 56	116 ± 102	101 ± 104
Treadmill					
Natural	413 ± 282	443 ± 288	211 ± 90	352 ± 178	461 ± 349
0.30 m/s	102 ± 77	207 ± 109	55 ± 49	71 ± 59	114 ± 96
0.20 m/s	111 ± 82	133 ± 88	70 ± 42	98 ± 83	109 ± 85

At slower speeds, EMG during both overground and treadmill trials altered in the same manner, both in reduction of EMG amplitude and selected alteration in timing. However, the MH appears to produce an additional burst at the stance-to-swing transition only at the slower walking speeds. These differences, listed in the following, represent the findings of EMG patterns that changed in timing and relative amplitude at the extremely slow walking speeds:

- TA maintained biphasic pattern but reduced in amplitude, particularly at early stance.
- MG maintained monophasic pattern but reduced in amplitude.
- RF lost the distinct peaks at the transitions from stance to swing and swing to stance; they were replaced by a low-level EMG pattern in loading.
- VL lost the peak at initial contact; it was replaced by a low-level EMG pattern in loading.
- MH showed a substantial reduction of EMG peaks at the transitions from swing to stance; they were replaced by activity during loading with an additional burst at toe-off.

DISCUSSION

We discuss the results of this study in relation to replication of past work on the differences between overground and treadmill gait. In addition, our trial revealed some findings at extremely slow walking speeds not reported in past work.

Normative kinematic and EMG gait patterns at natural speed, both overground and on treadmill, agree with past research [15–16]. Temporal-spatial results in the present study replicated past findings on the slight differences on a treadmill, including a slightly higher cadence and shorter stride length compared with overground [15–16,18]. Movement patterns at natural gait speed on overground were consistent with the literature [1–2,23–24] and the slight differences on the treadmill [15]. On the treadmill, average initial contact pattern was made with a full foot rather than initial heel contact [17], more flexed knee throughout stance [15], and more hip flexion in swing [15,18]. In the present trial, the trunk posture was slightly more flexed and the ankle showed less dorsiflexion in swing during the treadmill trials at natural speed. The trunk posture may be related to maintaining a forward position on the backward moving surface. The relatively reduced ankle dorsiflexion in swing may be related to the already enhanced hip flexion in swing, assuring foot clearance.

In the present study, muscle action showed some differences between overground and treadmill at natural speed, which agreed with Murray et al. [15] and Arsenault et al. [16]. Past research has indicated that treadmill gait generally promoted greater EMG amplitudes of all lower-limb muscles [15–16]. However, the present study only found increased EMG amplitudes in the distal muscles (TA and MG) at the natural speed on the treadmill compared with overground. In addition, the between-subject variability of kinematic and EMG variables were large both on the treadmill and overground at natural speeds. This observation contrasts earlier work that suggested a decrease in variability on the treadmill [16]. However, past work has also recognized that generally a large intersubject variability exists in normative gait studies [25].

The second component of the present study addressed the differences in kinematics and EMG at extremely slow walking speeds. As some minimal differences existed on the treadmill compared with overground at natural speed, these same differences were found at the slower walking speeds. The temporal-spatial parameters agreed with the literature that the cadence and stride decreased and the stance phase increased with a reduction of walking speed [15]. The treadmill results at slower speeds maintained a slight increase in cadence and less stride length compared with overground. A tendency for an increased variability in temporal-spatial results existed in the slower walking speeds compared with the natural speed.

The patterns of motion also agreed with past research, which mainly reported on speeds ranging from 0.60 to 0.80 m/s [9–10]. All segments decreased in motion—specifically, a foot-flat posture at initial contact, a loss of ankle plantar flexion at toe-off, a loss of knee flexion at weight acceptance, reduction of hip and knee flexion/extension, and a loss of forward trunk lean [10,13–14].

EMG results generally agree with the literature, demonstrating a substantial decrease in EMG amplitude of selected muscles with slower walking speed [9–11,13]. The phasic timing was also altered. The RF, VL, and MH muscles lost the distinct peaks of activity in early stance and mid to late swing. These muscles appeared to activate synchronously during loading at relatively lower EMG amplitude. The RF lost the peak at transition from stance to swing. which agrees with the literature at slower speeds [13,16]. Although a loss of EMG amplitude of the TA and MG muscles occurred, the proximal muscles showed a relatively greater reduction, which has been found in past studies [9,11]. The treadmill trials at extremely slow walking speeds appeared similar to all changes noted on overground.

In summary, at extremely slow walking speeds, both overground and treadmill, our results showed that—

- EMG amplitudes were substantially reduced from natural speed.
- EMG of proximal muscles were coactivating during loading in contrast to peaks at transition periods at natural speed.
- The TA and MG showed more similar timing to natural speed.

Some authors have argued that the decrease in joint ROM and EMG amplitude at the slower walking speeds is not simply a "gain" control of the neural control system [11]. In contrast, the observed changes could be predicted based on the change in functional requirements and mechanical demands at natural walking speed proposed by Winter [9] and Yang and Winter [11]. Walking demands can be categorized into the following: (1) control of the foot position, (2) energy absorption in early stance, (3) forward propulsion, (4) acceleration of the limb in early swing, and (5) deceleration of the limb in late swing. Functional requirements involve resistance to gravity and control of the speed of limb and body motion. These tasks are likely similar for overground and treadmill.

The muscle activity at the ankle performs two functions related to resistance to gravity during ambulation: (1) control of the foot position both at heel contact and swing (TA) and (2) forward propulsion (MG). At the slower walking speeds, the TA biphasic EMG pattern was intact because of the need to control the foot against gravity. Furthermore, the MG muscle maintained the same phasic pattern, indicating the continuing need for forward progression even at the slower walking speeds.

The hip and knee segments contribute to three demands of walking by (1) absorbing energy and lessening the impact at early stance and weight acceptance, (2) contributing to forward acceleration and foot clearance during swing, and (3) decelerating the lower limb in late swing [9]. At the two slower speeds of walking, shock absorption (at the knee) and acceleration and deceleration components of the hip and knee appear to be diminished, with an associated loss of hip and knee flexion at weight acceptance and decreased hip and knee flexion during the swing phase. A shortened stride length results from these alterations.

Therefore, the need to accelerate and decelerate the limb was minimized at the slower speeds, reflected by the altered timing of peak bursts and the lower EMG amplitudes of muscle activations, particularly of the proximal muscles. These findings agree with Yang and Winter [11] and Shiavi et al. [13].

The trunk segment revealed less trunk flexion at the slower speeds. The role of the trunk segment at natural speed has been suggested to (1) transfer and absorb the forces generated by the lower limbs to limit excessive vertical and lateral displacement of the head, (2) control gravitational force, (3) maintain equilibrium and balance within the base of support, and (4) contribute to the initiation and control of gait speed [1,26]. At the slower speeds, less muscle action of the lower limbs may be reflected in reduced trunk motion. Less acceleration and deceleration of the lower limbs would allow the body to remain within the base of support maintaining the chosen walking speed [26]. However, subjects often reported that balance was more difficult to maintain at the slower speeds [13]. The interaction of the trunk and lower limbs may be different at substantially slower walking speeds. Further biomechanical research is warranted as to the role of the trunk segment at these slow speeds.

OBSERVATIONS AND CLINICAL IMPLICATIONS

A loss of motion and slower walking speed is commonly seen in patients with neurological gait deficits [5,27–28]. A substantial decrease in walking speed may also induce a loss of automatism [13] and more precarious balance. Subjects in the present study reported that more conscious effort was required to maintain balance at these slower walking speeds. Patients with balance and sensorimotor deficits may be further challenged by the fact that walking speed is substantially slower. Thus, slow walking may be even more difficult for some patients than a more rapid gait.

For clinicians, findings from this study suggest that the effect of extremely slow walking speed would significantly reduce the ROM in most segments, particularly in the acceleration phase, of the gait cycle. EMG of muscle activation would be substantially reduced. Altered timing of muscle activation of the proximal muscles would be expected. The distal muscles would be expected to produce a more normal phasic pattern. Quantitative gait analysis would need to isolate the primary pathologies from the patterns induced by the slower walking speeds.

For clinical gait training, encouraging faster walking speeds within the security of a treadmill with or without weight support may be advantageous, as proposed by Behrman and Harkema [29] and Sullivan et al. [30]. Furthermore, the use of a treadmill in the evaluation of neurological patients would appear to simulate overground walking in the basic kinematics and EMG of muscle activity. The effect of the treadmill at the slower walking speeds would suggest to clinicians that encouragement should be given not only to increase walking speed but also to avoid trunk lean, promote more hip and knee extension in stance, increase stride length, and encourage heel strike at initial contact. This study would support these previously reported treatment targets [29–32]. Although this preliminary work is limited in sample size, the findings suggest that the effects of extremely slow walking speeds alter the normal gait pattern even more than results found in the past research at greater than 0.60 m/s [33]. The advantage of using a treadmill for more specific, immediate correction can be supported as long as clinicians are aware that some aspects may differ from the overground condition.

CONCLUSIONS

This study revealed minimal differences between overground and treadmill walking at natural speeds. At extremely slow walking speeds, overground and treadmill conditions showed similar and substantial reductions in kinematic and EMG patterns. The trunk and lower-limb ROMs were reduced, as well as EMG amplitudes of all muscles studied. Alterations in muscle timing appeared to be muscle-specific and may be related to the mechanical demands in walking at varying speeds. The main findings at the extremely slower walking speeds were a loss of both the generation and absorption of force function of most of the muscles studied. However, foot clearance, safe foot position at initial contact, and forward progression are maintained, regardless of the speed of walking or the use of a treadmill.

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