Biomechanics of overground vs. treadmill walking in healthy individuals

Song Joo Lee¹,² and Joseph Hidler¹,²

¹Department of Biomedical Engineering, Catholic University, and ²Center for Applied Biomechanics and Rehabilitation Research National Rehabilitation Hospital, Washington, District of Columbia

Submitted 5 December 2006; accepted in final form 21 November 2007

Lee SJ, Hidler J. Biomechanics of overground vs. treadmill walking in healthy individuals. J Appl Physiol 104: 747–755, 2008. First published November 29, 2007; doi:10.1152/japplphysiol.01380.2006.—The goal of this study was to compare treadmill walking with overground walking in healthy subjects with no known gait disorders. Nineteen subjects were tested, where each subject walked on a split-belt instrumented treadmill as well as over a smooth, flat surface. Comparisons between walking conditions were made for temporal gait parameters such as step length and cadence, leg kinematics, joint moments and powers, and muscle activity. Overall, very few differences were found in temporal gait parameters or leg kinematics between treadmill and overground walking. Conversely, sagittal plane joint moments were found to be quite different, where during treadmill walking trials, subjects demonstrated less dorsiflexion moments, less knee extensor moments, and greater hip extensor moments. Joint powers in the sagittal plane were found to be similar at the ankle but quite different at the knee and hip joints. Differences in muscle activity were observed between the two walking modalities, particularly in the tibialis anterior throughout stance, and in the hamstrings, vastus medialis and adductor longus during swing. While differences were observed in muscle activation patterns, joint moments and joint powers between the two walking modalities, the overall patterns in these behaviors were quite similar. From a therapeutic perspective, this suggests that training individuals with neurological injuries on a treadmill appears to be justified.

OVER THE LAST DECADE, THERE has been a steady shift in gait training strategies in neurorehabilitation clinics where body-weight-supported treadmill training is now considered a viable intervention for treating gait impairments following neurological disorders such as stroke (12, 13) and spinal cord injury (9). Treadmill training has numerous advantages compared with overground gait training, where the training can be done in small area, a larger volume of steps can be achieved, walking speed can be well controlled, and because the subject is stationary and often elevated, the positioning of the therapist is more optimal for providing assistance. Additionally, overhead body weight-support systems can be utilized, relieving the subject of a portion of their weight, which allows them to train safely and earlier in their recovery period (see Ref. 10 for a review).

Because the goal of all patients is to walk overground and not on a treadmill, it is important that the motor control strategy utilized during each type of walking modality be similar so that improvements in treadmill ambulation will transfer to overground walking. In theory, if the belt speed of the treadmill is constant, biomechanically, there should be no differences between the two walking modalities (31). A number of previous studies have compared temporal gait parameters (1, 5, 24–26, 29, 33), joint kinematics (1, 5, 24, 25, 29), and muscle activation patterns (2, 24, 25) between overground and treadmill walking, however, the results are often conflicting and inconclusive. For example, Murray et al. (24) reported no statistical differences in temporal gait parameters but claimed that during treadmill walking, subjects demonstrated trends for shorter step lengths, higher cadences, shorter swing phases, and longer double-limb support. They also reported that despite no statistical differences in muscle activity except for quadriceps, EMG activity was, in general, higher on the treadmill. This is in contrast to the reports by Arsenault et al. (2), who reported no differences in muscle activity in the soleus, rectus femoris, vastus medialis, or tibialis anterior between the two walking modalities. These investigators did report elevated activity in the biceps femoris and the variability was lower in muscle firing patterns on the treadmill. Alton et al. (1) found that during treadmill ambulation, subjects had a shorter stance time, higher cadence, larger hip range of motion, and greater maximum hip flexion angle. Unfortunately, the experimental procedures in that study were questionable, because the walking speeds between the two conditions were never accurately matched, and the method for estimating the joint angles, particularly at the hip, was problematic.

Although the studies described above have looked at temporal gait parameters, kinematics, and muscle activation patterns, only one has looked at joint moments and powers while walking on a treadmill (29). In that study, it was reported that joint moments and powers were statistically different, yet because these differences were within the variability of the kinetic measures, it was concluded that treadmill and overground joint moments and powers are similar. This study only reported on the maximum and minimum moments and powers during stance, and it did not report on muscle activation patterns for these trials.

Although the study by Riley et al. (29) provides the first look at kinetic patterns during treadmill walking, it is important to look at these behaviors across the gait cycle and also to examine the corresponding muscle activation patterns. Because motor coordination, synergy patterns, energy expenditure, and other control strategies can be best observed by looking at kinetics and muscle activity, having a detailed understanding of these behaviors for treadmill and overground gait may provide important insight into developing gait training protocols for neurological subjects and for understanding how treadmill-based interventions may be limited. It may also help explain the therapeutic benefits of body weight-supported treadmill training for individuals with lower limb impairments.

Address for reprint requests and other correspondence: J. Hidler, Dept. of Biomedical Engineering, Catholic Univ., Pangborn Hall, #104b, 620 Michigan Ave., NE, Washington, DC 20064 (e-mail: hidler@cua.edu).

http://www.jap.org

8750-7587/08 $8.00 Copyright © 2008 the American Physiological Society

The costs of publication of this article were defrayed in part by the payment of page charges. The article must therefore be hereby marked “advertisement” in accordance with 18 U.S.C. Section 1734 solely to indicate this fact.
With the integration of force-sensing capabilities into split-belt treadmills (3, 22), ground reaction forces and centers of pressure can now be measured so that when combined with kinematic data, inverse dynamics calculations of joint moments and powers during treadmill ambulation can now be realized.

The goal of this paper is to provide a comprehensive analysis of the temporal gait parameters, joint kinematics, joint kinetics, and muscle activation patterns utilized when subjects walk on a treadmill compared with overground ambulation. We hypothesize that when individuals walk on a treadmill, they will demonstrate significant differences in joint moments and powers, as well as muscle activation patterns when compared with overground walking.

METHODS

Subjects

Nineteen healthy individuals with no known gait impairments participated in the study. Eight subjects (4 men, 4 women) were in the age range of 50–70 yr old (means ± SD: 56.0 ± 5.6), while 11 subjects (5 men, 6 women) were between 18 and 30 yr old (means ± SD: 23.5 ± 3.3 yr). Our original goal was to determine whether there were age-related differences in treadmill vs. overground walking. However, with analysis of the data, we found that there were no age-related differences in any of the metrics we examined. As such, we collapsed all subject data into the analysis.

Exclusion criteria included cardiac arrhythmia, hypertension, or any known gait abnormality such as an orthopedic injury, lower limb pain, or neurological injury that would bias the results of this study. All subjects were required to provide informed consent approved by the Institutional Review Boards of Medstar Research Institute. All experiments were conducted at the National Rehabilitation Hospital (Washington, DC).

Instrumentation

The primary equipment used in this study was an ADAL3D-F/COP/Mz instrumented split-belt treadmill (TECHMACHINE, Andrézieux, France; see Ref. 3 for detailed description). As shown in Fig. 1, each half of the treadmill is mounted on 4 Kistler triaxial piezoelectric sensors (Winterthur, Switzerland). This allows for ground reaction forces to be resolved in the vertical, anterior-posterior, and medial-lateral planes while subjects ambulate on the device. These forces could then be used to calculate the center of pressure under each foot during stance.

Compared with previous studies that compared treadmill with overground walking (1, 2, 5, 24–26, 33), one of the distinct advantages of our experimental setup is that the floor in our laboratory is raised to be level with the ADAL treadmill. Thus, for the overground experiments, the treadmill motors were turned off and the subjects simply walked across the treadmill. Utilizing the same “force plates” for both treadmill and overground walking trials eliminated any experimental bias that might have been introduced into the ground reaction force measurements had the overground gait analysis been done using standard force plates. In that setting, the sensors of the different force plates may have different dynamic response characteristics. More importantly, the surfaces of the force plates in the two walking modalities would be different (e.g., rubber treadmill vs. hard overground). Changes in surface stiffness and damping result in significant changes in muscle EMG activity (23), suggesting the importance of consistent test conditions. It should be noted that before the study, we checked for belt slippage during overground walking trials by placing motion analysis markers on the surface of the belt as subjects walked across the treadmill. We did not observe any belt slippage during these overground test trials.

The kinematics of lower limbs were measured using a CodaMotion system (Charnwood Dynamics). A single Codemotion CX1048 infrared camera station was used to capture the three-dimensional coordinates of markers on the subject’s lower limbs. To minimize marker movement artifacts that come from direct contact to subjects’ skin, custom-made clusters were used to track the markers. Each cluster consisted of four Codemotion active markers attached to custom-made Aquaplast shells, where the marker locations could be adjusted for each subject to minimize marker dropout.

Muscle activity was recorded differentially from the tibialis anterior, medial gastrocnemius, medial hamstrings, vastus medialis, rectus...
femoris, adductor longus, and gluteus maximus and medius muscles on the left leg using a Bagnoli-8 EMG system (Delsys, Boston, MA). All force and EMG data were antialias filtered at 500 Hz before sampling at 1,000 Hz using a 16-bit data acquisition board (Measurement computing, PCI-DAS 6402, Middleboro, MA) and custom software (Mathworks, Natick, MA). Marker position data from C-DMotion were sampled at 100 Hz. Force plate data were further low-pass filtered using a zero-delay fourth-order Butterworth filter with a 25-Hz cutoff frequency.

Protocol

After the subject signed the informed consent form, EMG electrodes were attached on the subject’s skin by a trained physical therapist, after which the marker clusters were strapped on to the subject’s legs with neoprene (DuPont, Wilmington, DE) and coband (3M, St. Paul, MN). The neoprene was wrapped tightly around the subject’s shanks and thighs, which helped minimize skin movement artifacts. Marker clusters were placed on the subject’s feet, shanks and thighs, while four individual markers were placed on the subject’s pelvis.

With all of the instrumentation in place, subjects were asked to walk overground for a distance of ~5 m at their comfortable speed. In general, ~10 trials were necessary to obtain three acceptable passes with adequate rest breaks in between. The criteria for an acceptable trial were that the subject made good contact with the center of the force plate, and minimal marker dropout was observed over a full gait cycle. The average walking speeds were calculated from the first three overground trials, which were then used in the treadmill trials.

After the overground walking trials, subjects were allowed to acclimate to walking on the treadmill for ~3 min. During the treadmill trials, subjects were not allowed to hold onto the handrails because this could possibly alter their gait pattern. After the acclimation phase, the speed of the treadmill was set to the speed obtained during the overground walking trials and data were collected for 30 s.

On completion of the walking trials, anatomic landmarks were digitized using the C-Motion digitizing pointer, which established relationships between landmarks and marker locations. Landmark locations were obtained from both legs for the lateral malleolus, lateral femoral epicondyle, greater trochanter, and the anterior iliac crest. Additionally, anatomic measures of the foot, ankle, and knee were taken for each subject to create subject-specific link segment models in Visual 3D (C-Motion, Rockville, MD) (see below).

Analysis

Subject-specific models were created in Visual 3D using the anatomic measures taken for each subject, along with body mass and

Fig. 2. Average ankle, knee, and hip joint moments in sagittal (A, C, E) and frontal (B, D, F) planes for all subjects. Gray line with shading, mean ±1 SD for overground trials; solid black line with 2 dashed lines, mean ±1 SD for treadmill trials. Max Pflexor, maximum plantar flexion; Max Dflexor, maximum dorsiflexion; MaxEV1, maximum eversion in early stance; MaxEV2, maximum eversion in late stance; MaxIN, maximum inversion; MaxEX1, maximum extension in early stance; MaxEX2, maximum extension in late stance; MaxFL, maximum flexion; MaxFL1, maximum flexion in mid stance; MaxFL2, maximum flexion in late swing; MaxIN, maximum inversion; MinVal, minimum valgus; MaxVal1, maximum valgus in early stance; MaxVal2, maximum valgus in late stance; MinAB, minimum abduction; MaxAB1, maximum abduction in early stance; MaxAB2, maximum abduction in late stance.

J Appl Physiol • VOL 104 • MARCH 2008 • www.jap.org
height, which were used to normalize joint moments and joint powers, and to define segment masses and inertial values.

The Visual 3D model assumes that each segment is a rigid body, which is then used to calculate joint angles, joint moments, and joint powers.

For each trial, individual gait cycles were extracted from a data sequence, considered the interval between successive heel strikes in the same foot. Since only one heel contact was normally made on the force plate for the overground trials, the subsequent heel strike was inferred from the kinematic data. Each stride cycle was resampled for averaging purposes and time normalized, expressed as a percentage of the total gait cycle (e.g., 0–100%). All temporal, kinematic, kinetic, and EMG measures were calculated for each stride, which resulted in three overground trials and ~10–20 treadmill strides. To make equivalent comparisons between the two walking conditions, 3 strides were randomly selected from the treadmill trials for each subject, which were then compared with the overground trials (see Statistics).

Temporal gait parameters. Temporal gait parameters were estimated for treadmill and overground walking conditions. Measures included stance time, total double-limb support time, swing time, stride time, step time, cadence, and stride length (see Ref. 21 for an explanation of each gait parameter). Speed, cadence, and stride length were normalized using the procedures described by Hof (15) to account for individuals of various heights:

Normalized speed: \[ \text{speed} \cdot \frac{1}{\sqrt{H \cdot g}} \]  
(1)  

Normalized cadence: \[ \text{cadence} \cdot \frac{H}{g} \]  
(2)  

Normalized stride length: \[ \text{stride length} \cdot \frac{1}{H} \]  
(3)

where \( H \) is the subject’s height in meters, and \( g \) is gravity (9.81 m/s²).

Kinematic analysis. The angular range of motion in the sagittal plane for the hip, knee, and ankle was found by subtracting the minimum joint angle from the maximum joint angle for overground and treadmill trials (1). Peak ankle flexion and extension, knee flexion, and hip flexion and extension were also identified for each trial. These values, and the phase of the gait cycle in which they occurred, were used to evaluate the kinematic differences between treadmill and overground walking.

Kinetic analysis. Joint moments at the ankle, knee, and hip (11, 21, 34, 35) were analyzed in the sagittal and frontal planes. For the ankle, the maximum dorsiflexor and plantar flexor moments were calculated in the sagittal plane, while maximum eversion and the midstance minimum moments were determined for the frontal plane. For the knee in the sagittal plane, the maximum extension moments generated in early stance and late stance, along with the maximum flexion moment were identified. In the frontal plane, the two peak valgus moments in early and late stance were identified, along with the minimum valgus moment in midstance. At the hip, sagittal plane maximum extension moments were identified in early stance and late swing, along with the maximum hip flexion moment. For the frontal plane, peak abduction moments in early and late stance were identified as well as the minimum in midstance. These metrics are outlined in Fig. 2 and described in RESULTS.

Joint powers (11) were analyzed for the sagittal planes, where the minimum and maximum joint powers were estimated for the ankle, knee, and hip throughout the gait cycle (Fig. 3). Ground reaction forces were examined in the vertical, medial-lateral, and anterior-posterior planes, where the maximum and minimum values were analyzed in the gait cycle, particularly at transition points (see Fig. 4).

EMG analysis. Muscle activation patterns (16) were processed using the technique described by Hidler and Wall (14). Briefly, the mean EMG pattern for each muscle was rectified and then smoothed using a 50-point root-mean-square algorithm (19). For each muscle, the smoothed EMG was normalized to the maximum value observed in that respective muscle across all trials so that intersubject comparisons could be made. After the average EMG profile for each muscle was calculated, the data were broken up into seven phases as described by Perry (27). Within each of these phases, the integrated EMG activity was calculated for each muscle.

Statistics

Two parallel statistical analyses were performed on the temporal gait parameters, kinematic measures, joint moments, and joint powers. A repeated-measures ANOVA was used to first compare treadmill and overground walking. As stated above, the three overground strides were compared with three randomly selected treadmill strides for each subject. Temporal parameters compared included normalized walking speed, normalized cadence, normalized stride length, stride time, stance time, swing time, and double-limb support time. Kinematic variables examined were maximum and minimum flexion and extension angles, and range of motion of the ankle, knee, and hip in sagittal and frontal plane. Kinetic measures were compared by looking at anterior-posterior, medial-lateral, and vertical ground reaction forces, joint moments, and joint powers that have been discussed in Kinematic analysis. Interactions were also looked at across strides and groups, as well as within groups and between strides. All statistical tests were run using STATA (Intercooled Stats 9.2 for Windows, Stata).
Because of the large number of comparisons being made between treadmill and overground walking, we also analyzed the data described above using a seemingly unrelated regression (SUR) model, which is an extension of linear regression model. SUR models are systems of simultaneous equations in which the variables are not independent. The SUR model tests the effects of a number of independent variables on each dependent variable by taking into account the potential correlation among the error terms. Thus this modeling technique allowed us to look for differences between treadmill and overground walking by implicitly modeling the similarities between the dependent variables (20). For the SUR model analysis, independent variables consisted of type of walking and stride number.

To analyze the EMG data, we used a fixed-effects regression analysis to account for the interrelatedness between the phases of the gait cycle and to model the unobserved factors that arise from the multiple steps and the individual subjects. This model allowed us to test for differences between the treadmill and the overground for each phase of the gait cycle, assuming that the phases are part of a complete movement rather than analyzing each phase separately across the types of walking. Here, fixed factors in the model included type of walking, subject, and phase of the gait cycle, while the dependent measure was the mean integrated amount of muscle activity (see EMG Analysis).

RESULTS

Temporal Gait Parameters and Joint Kinematics

Table 1 lists means SD of the temporal parameters for both overground and treadmill ambulation. With the exception of swing time and stance time, none of the temporal gait parameters were significantly different between treadmill and overground ambulation. Comparing joint kinematics in the sagittal plane, only knee range of motion was significantly different between treadmill and overground walking. A summary of joint kinematic parameters is listed in Table 2.

Joint Moments

General observation. With the exception of peak ankle plantar flexion, sagittal plane joint moments during treadmill trials were significantly different than those utilized during overground walking. Conversely, joint moments in the frontal plane were not significantly different between treadmill and overground walking. A detailed listing of the mean and standard deviations of the sagittal plane joint moment features outlined in Kinetic analysis can be found in Table 3.

Ankle moment. A representative example of the ankle moments for treadmill and overground walking can be seen in Fig. 2A. During the loading response of the gait cycle (0–10%), subjects had a tendency to produce larger dorsiflexor moments during overground walking. Maximum plantar flexor moments during terminal stance phase (30–50%) were not significantly different between overground and treadmill walking. For the frontal plane, neither ankle inversion nor eversion moments were significantly different between the walking conditions (Fig. 2B).

Table 1. Temporal gait parameters for overground and treadmill walking

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Overground</th>
<th>Treadmill</th>
<th>P Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking speed (dimensionless)</td>
<td>0.27 (0.04)</td>
<td>0.28 (0.04)</td>
<td>0.95</td>
</tr>
<tr>
<td>Step time, s</td>
<td>0.56 (0.05)</td>
<td>0.54 (0.06)</td>
<td>0.41</td>
</tr>
<tr>
<td>Stance time, s</td>
<td>0.68 (0.07)</td>
<td>0.65 (0.08)*</td>
<td>0.021</td>
</tr>
<tr>
<td>Swing time, s</td>
<td>0.45 (0.03)</td>
<td>0.43 (0.04)*</td>
<td>0.0017</td>
</tr>
<tr>
<td>Double-limb support time, s</td>
<td>0.23 (0.05)</td>
<td>0.22 (0.05)</td>
<td>0.77</td>
</tr>
<tr>
<td>Cadence (dimensionless)</td>
<td>45.1 (4.0)</td>
<td>46.6 (4.5)</td>
<td>0.28</td>
</tr>
<tr>
<td>Stride length (dimensionless)</td>
<td>0.73 (0.09)</td>
<td>0.71 (0.08)</td>
<td>0.86</td>
</tr>
</tbody>
</table>

Values are means (SD). *Different from overground walking, P < 0.05 (see Statistics for details).
Knee moment. Throughout the gait cycle, subjects produced larger knee extensor moments during overground walking compared with treadmill walking. As shown in Fig. 2C and summarized in Table 3, the peak extensor moments in early and late stance (e.g., MaxEX1 and MaxEX2) were both significantly greater during overground walking than on the treadmill ($P < 0.05$). The peak flexor moments in late stance (e.g., MaxFL1) and in late swing (e.g., MaxFL2) were significantly greater during treadmill walking. In the frontal plane (Fig. 2D), no statistical differences were found in valgus knee moments between walking conditions.

Hip moment. Sagittal plane hip extensor moments in early stance (e.g., MaxEX1) and late swing (MaxEX2) were both greater during treadmill walking than during overground trials (see Fig. 2E), whereas the peak flexion moment (MaxFL) was significantly higher (e.g., more negative) during treadmill walking. In the frontal plane, no statistical differences were found in joint moments for treadmill or overground walking (Fig. 2F).

Joint Powers

Joint powers for the sagittal plane are reported. As illustrated in Fig. 3A and listed in Table 4, the maximum and minimum joint powers at the ankle for treadmill and overground trials were not different. Conversely, significant differences were found for knee power curves throughout the gait cycle. Figure 3B illustrates the mean knee power curves for both treadmill and overground walking, along with the metrics used for comparison. The first two local minima (e.g., Min1 and Min2) were found to be significantly less for overground walking (e.g., more negative), whereas the third local minimum (Min3) was not different for the two walking conditions. The maximum knee power (Max) was also not significantly different between treadmill and overground walking. For the hip, the first maximum (Max1) was significantly greater for treadmill walking while the local minimum (Min) was significantly greater (e.g., more negative) for overground walking. No differences were found for the second maximum hip power (Max2) between walking conditions.

Ground Reaction Forces

Because the joint kinematics were similar between treadmill and overground walking yet numerous joint moments and powers were different, we examined the ground reaction forces for each condition in hopes of better understanding these differences. Figure 4 illustrates the ground reaction forces in the anterior-posterior (A), medial-lateral (B), and vertical planes (C). We found that the maximum braking forces (labeled MinAP in Fig. 4A) was significantly greater during overground walking than during treadmill walking. No other ground reaction force metrics shown in Fig. 4 were significantly different.

In addition to comparing absolute measures of ground reaction forces, we also examined the magnitude of the various ground reaction forces at the time when metrics used to describe sagittal plane joint moments were observed (e.g., those listed in Table 3). When peak dorsiflexor moments were observed, the braking forces and the vertical ground reaction forces were significantly higher during overground walking ($P < 0.05$). We also observed that the propulsion forces were significantly lower during overground walking at knee MaxFL1 and significantly higher during overground walking at knee MaxEX2 ($P < 0.05$).

Muscle Activity

During treadmill walking, EMG activity in the tibialis anterior was lower throughout stance ($P < 0.05$). Similarly, activity in the gastrocnemius was also lower throughout much of stance yet slightly higher in terminal swing ($P < 0.05$). An interesting pattern among hamstrings, vastus medialis, and adductor longus emerged between the two conditions. Here,
throughout early and midswing, there was higher activity during overground walking in each of these muscles yet at terminal swing, this relationship reversed (e.g., significantly more activity during treadmill walking). For the rectus femoris, there was significantly higher muscle activity when subjects walked on the treadmill during the transition from stance to swing (e.g., phases 4 and 5) as well as at terminal swing. Table 5 summarizes integrated muscle activation parameters for treadmill and overground walking conditions.

**DISCUSSION**

Although numerous studies have compared treadmill and overground walking, there still exists significant debate as to the differences between the two walking modalities. Our results suggest that when individuals walk on a treadmill, they modify their muscle activation patterns and subsequently joint moments and powers while maintaining relatively constant limb kinematics and spatiotemporal gait parameters.

We did not find any statistical differences in the peak vertical ground reaction forces between treadmill and overground walking. This is different from the reports by White et al. (33), who found that for normal and fast walking speeds, the magnitude of the vertical ground reaction forces were greater on the treadmill during midstance yet lower in late stance. Because those authors evaluated the same events as we did in this study (e.g., VMax1, VMin, and VMax2 in Fig. 4C), one possible explanation for the differences in results is that they collected their overground data on a different system than the treadmill data. We believe that a key strength of this study was that both overground and treadmill data were collected from the same force plates, which were mounted underneath each half of the treadmill. This resulted in consistent sensors being used for both walking modalities, as well as the same surface, both of which could introduce errors into the recordings.

We also did not find significant differences in medial-lateral ground reaction forces. Riley et al. (29) reported significantly different medial-lateral forces at self-selected speeds, which led to statistical difference in frontal plane joint moments. They claimed that the differences in these forces were in the range of variability. Again, in that study, overground gait assessment was done in a motion analysis laboratory rather than on the treadmill. Because the surfaces of the force plates were different for the two conditions in their study, whereas in this study we had a consistent and continuous force plate surface, this may explain the differences with the results presented in this study.

When subjects walked on the treadmill, we did find the braking ground reaction forces at heel contact were less than overground walking (Fig. 4, MinAP), which led to smaller ankle dorsiflexor moments and smaller knee extensor moments. One possible explanation for this would be if the

Table 5. Means and standard deviations of EMG integration per each phase of gait cycle during overground and treadmill walking

<table>
<thead>
<tr>
<th>Phase</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Gastrocnemius</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibialis anterior</td>
<td>1</td>
<td>141.9 (56.7)</td>
<td>105.3 (27.2)*</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>74.2 (31.9)</td>
<td>66.1 (38.8)*</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>58.6 (19.1)</td>
<td>47.2 (15.6)*</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>41.8 (18.1)</td>
<td>29.3 (22.3)*</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>132.6 (32.7)</td>
<td>107.1 (24.5)*</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>57.6 (18.2)</td>
<td>66.0 (19.5)</td>
</tr>
<tr>
<td></td>
<td>7</td>
<td>162.0 (59.7)</td>
<td>154.9 (46.5)</td>
</tr>
</tbody>
</table>

| Hamstrings | 1 | 63.4 (34.9) | 44.6 (30.4)* | 1 | 119.0 (56.5) |
|           | 2 | 47.5 (23.9) | 48.5 (17.8) | 2 | 87.4 (53.3) |
|           | 3 | 65.8 (43.8) | 44.6 (8.6) | 3 | 46.8 (21.4) |
|           | 4 | 24.5 (27.3) | 19.9 (12.1) | 4 | 26.4 (14.8) |
|           | 5 | 23.0 (18.6) | 15.2 (7.1)* | 5 | 24.9 (10.8) |
|           | 6 | 46.7 (18.9) | 36.8 (15.1)* | 6 | 27.3 (16.1) |
|           | 7 | 105.5 (17.4) | 123.9 (24.4)* | 7 | 65.1 (32.2) |

| Rectus femoris | 1 | 111.5 (78.5) | 120.8 (58.1) | 1 | 128.4 (50.5) |
|               | 2 | 100.5 (86.5) | 93.2 (73.1) | 2 | 84.0 (38.1) |
|               | 3 | 54.8 (49.6) | 52.2 (45.5) | 3 | 60.6 (37.8) |
|               | 4 | 37.1 (20.7) | 47.6 (25.3)* | 4 | 32.6 (16.0) |
|               | 5 | 39.4 (20.2) | 66.0 (40.2)* | 5 | 39.4 (17.6) |
|               | 6 | 26.9 (20.1) | 27.2 (18.9) | 6 | 30.9 (17.6) |
|               | 7 | 56.7 (46.7) | 72.9 (39.5)* | 7 | 110.9 (33.7) |

| Gluteus medius | 1 | 120.7 (38.2) | 119.1 (22.9) | 1 | 111.4 (87.6) |
|                | 2 | 86.6 (36.5) | 94.5 (44.8) | 2 | 85.9 (38.1) |
|                | 3 | 52.7 (37.0) | 48.4 (25.4) | 3 | 73.7 (61.7) |
|                | 4 | 25.4 (18.0) | 24.2 (16.7)* | 4 | 67.0 (36.8) |
|                | 5 | 25.7 (17.0) | 28.1 (19.6)* | 5 | 125.7 (116.5) |
|                | 6 | 25.6 (18.0) | 25.5 (18.5) | 6 | 86.2 (47.5) |
|                | 7 | 87.0 (31.5) | 79.1 (22.1) | 7 | 127.1 (63.6) |

Values are means (SD). I, loading response (0–10%); 2, midstance (10–30%); 3, terminal stance (30–50%); 4, preswing (50–60%); 5, initial swing (60–75%); 6, midswing (73–85%); 7, terminal swing (85–100%); Different from overground walking, P < 0.05 (see Statistics for details).
treadmill belt slowed briefly at heel contact due to excessive loads placed on the treadmill motors. In looking at the output of the treadmill motor encoders, we did observe speed decreases that reached a maximum of 2.5% of the reference speed. Although this speed fluctuation may appear to be small, it could result in the slight decreases in braking forces observed in the study. In addition, this speed fluctuation could also be responsible for the attenuation in vertical ground reaction forces when subjects produced a maximum dorsiflexor moment.

In contrast to Riley et al. (29), our results demonstrate that there are in fact significant differences in sagittal plane joint moments and powers between treadmill and overground walking. They found 15 of 18 joint moment maximums and 3 of 6 power maximums, each in stance, were different but concluded that because the differences were in the range of repeatability of kinematic measures, the differences should not be considered meaningful. We too found that there was significant variability in joint moments; however, many of the metrics we compared (e.g., see Figs. 2 and 3) were still significantly different. In fact, these differences in joint moments and powers are further supported by the differences we observed in muscle activation patterns in numerous muscles (see Statistics).

Our original aim was to determine whether there are differences between walking on a treadmill compared with walking overground, in terms of kinematics, kinetics, and muscle activation patterns. While we found that there were differences in joint moments, joint powers, and EMGs, the key question now is why are these behaviors different? Based on the study by Van Ingen Schenau (31), if the treadmill belt speed remains constant, there should, in theory, be no differences in the dynamic behavior between the two conditions. We believe that there are a number of factors that probably violate this theoretical law. First, most studies have compared walking on the treadmill with overground walking using different force plates (e.g., the force plates under the treadmill and the gait laboratory force plate). This could introduce mechanical errors that the subject may need to compensate for. However, in our study, this was controlled for because the same surface was used in both walking conditions. Second, if the treadmill belt speed does in fact change, then the acceleration patterns of the limbs under the two conditions would change, requiring a different control strategy to maintain constant kinematics. Similar to previous studies (29, 30), we did observe a slight drop in belt speed at heel contact (e.g., <2.5%) which can explain the differences in the dorsiflexor moments we observed in early stance. However, the differences in knee and hip moments observed later in the gait cycle would presumably not be affected by this issue.

Perhaps the most likely reason we observed differences between the two walking conditions is that optic flow subjects receive on the treadmill is in stark contrast to what they receive while walking overground. Studies have shown that vision plays a role in gait (32) such that different optic flow patterns may alter locomotor control strategies. When walking on the treadmill, subjects do not receive the same optic flow as they do when walking overground, which may alter their balance and stability or their perception of where they are on the treadmill or the speed at which they are ambulating. These factors are supported by previous studies, because Regnaux et al. (28) found that treadmill ambulation was not an automated task. Since the optic flow patterns are different between treadmill and overground walking, small perceptual changes at one stage of the gait cycle could lead to altered joint moments and consequently to a cascade of changes during the remainder of the gait cycle if the goal is to preserve their kinematic patterns, which is what we hypothesize. This hypothesis is further supported by Carollo and Matthews (6) and Ivanenko et al. (17, 18), who suggested that during human locomotion, the kinematics appear to be the desired control variable. Satisfying kinematic demands during functional tasks has also been reported in primates (4, 7).

Our findings suggest that while temporal gait parameters and kinematic patterns are similar between treadmill and overground walking, the muscle activation patterns and joint moments and powers used to achieve these movement patterns are often different. Although such effects may lead to differences in motor strategies, from a therapeutic perspective, the overall kinematic and muscle activation patterns appear to be similar enough that training individuals with neurological injuries (e.g., stroke and spinal cord injury) on a treadmill appears justified. Because walking at home and society often requires individuals to navigate obstacles and alter their strategies, it could be viewed that the slight differences we observed here could be beneficial to the individual being able to adapt to different environments.

ACKNOWLEDGMENTS

We thank Mihrkiye Mete for helping with statistical analysis, Scott Selbie at C-Motion for suggesting a visual 3D model and all subjects who volunteered for the study.

REFERENCES


