ORIGINAL ARTICLE

Effects of Treadmill Inclination on the Gait of Individuals With Chronic Hemiparesis

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Objective: To analyze the effects of electric treadmill inclination on the gait of individuals with chronic hemiparesis.

Design: Descriptive, observational study.

Setting: Laboratory for human movement analyses of UFRN.

Participants: Individuals (N=18) with a mean age of 55.3±9.3 years and a lesion time of 36±22.8 months.

Interventions: Not applicable.

Main Outcome Measures: Subjects were assessed for functional independence (FIM) and balance (Berg Balance Scale).

Spatial-temporal variables were observed as well as the angular variation of the hip, knee, and ankle in the sagittal plane, while the individuals walked on the treadmill at 3 different inclinations (0%, 5%, and 10%).

Results: There was an increase in stance time between 0% and 5% (0.83±0.21 vs 0.87±0.20; P=.011) and 0% and 10% (0.83±0.21 vs 0.88±0.23; P=.021). The other spatial-temporal variables did not change. During initial contact there was an increase in the hip, knee, and ankle flexion angle. An increase in hip amplitude was also observed between 0% and 10% (38.80±5.96 vs 41.12±5.63; P=.002) and in knee amplitude between 0% and 10% (47.51±15.07 vs 50.30±12.82; P=.040), as well as decreased hip extension and increased dorsiflexion.

Conclusions: Treadmill inclination promoted angular alterations such as an increase in hip, knee, and ankle angle during initial contact and the swing phase and an increase in the amplitude of movement of the hip and knee, as well as an increase in stance time of the paretic lower limb.

Key Words: Biomechanics; Exercise test; Rehabilitation; Stroke.

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inclined surfaces produces alterations in the kinematic patterns and spatial-temporal variables of hemiparetic individuals.

METHODS

Participants

The sample was composed of individuals with chronic hemiparesis after ischemic or hemorrhagic stroke and who were undergoing physical therapy treatment. The inclusion criteria were as follows: spasticity classified between levels 0 and 2 on the Modified Ashworth Scale of muscle spasticity\(^34\) for the lower limb affected; ambulatory capacity classified between levels 3 and 5 on the Functional Ambulatory Classification\(^27\); minimum sequela time of 6 months; absence of clinical signs of cardiac alterations (New York Heart Association, degree I\(^28\); absence of other orthopedic or neurologic impairment that caused gait alterations; not using orthotics on the paretic lower limb; and capacity to obey simple verbal commands. Exclusion criteria were: individuals whose systolic blood pressure rose 10mmHg during the treadmill test\(^29\) or whose heart rate exceeded 75% of the age-adjusted maximum heart rate according to the formula proposed by Tanaka et al.,\(^30\) and those who had a fear of falling while walking on the treadmill. All the individuals signed an informed consent form, and the study was approved by the institutional research ethics committee.

Measuring Instruments

Functional independence in the activities of daily living was assessed using the FIM.\(^31\) To evaluate balance, we used the Berg Balance Scale,\(^32\) which has proved to be an effective instrument for assessing the balance of hemiparetic individuals.\(^33-35\)

Gait analysis was conducted with the subjects walking on a Gait Trainer System 2 treadmill\(^6\). A Biodex Unweighting System\(^2\) was used, but no weight was supported by the system. It was used for safety purposes only, to avoid falls or false steps. The data were captured using the Qualisys Motion Capture System\(^9\). Three Qualisys ProReflex MCU-240 cameras that emit infrared light and spherical passive markers were used. The data were captured at a frequency of 120Hz.

Experimental Protocol

After initial assessment, where clinical and demographic data, anthropometric measures, and vital signs (heart rate and blood pressure) were collected; functional independence and balance assessments were made to characterize the sample. We conducted gait analysis with barefoot individuals, placing reflective markers on the following bone structures: greater trochanter, medial and lateral epicondyle of the femur, medial and lateral malleolus, calcaneus, and the 5th and 1st metatarsal head. Markers were used to trace the segments in space during movement and were placed on the middle third and lateral surface of the thigh and leg. These consisted of 4 markers noncollinearly placed on a square base fixed with Velcro to an elastic neoprene strip. Markers were also placed on the calcaneus and 1st metatarsal head of the nonparetic foot.\(^36-38\)

The speed used on the treadmill was calculated using a chronometer during overground walking on a 10-m long walkway.\(^16\) The mean velocity obtained from the 3 passes on the walkway was used, disregarding the acceleration and deceleration phases. Later, the individuals were instructed on the correct trunk posture and upper limb position and transfer weight over the hemiplegic limb to be used during the treadmill protocol. All the individuals were capable of understanding this command and walked satisfactorily on the treadmill.

Static collection was conducted with the individuals in the orthostatic position (position of reference) in order to provide data for the future creation of their biomechanical model. The data were collected for 5 seconds.

The treadmill was then turned on and the speed gradually increased until it reached the velocity calculated during the 10-m test. After 2 minutes, a 30-second video capture was made (dynamic collection). All the individuals were submitted to collections on the treadmill with inclinations of 0%, 5%, and 10%, according to earlier studies.\(^10,12\) To avoid fatigue, a 3-minute rest period was given between the different conditions.\(^16\)

Data Analysis

The data generated by Qualisys were exported to Visual 3D version Basic/RT 3.99.25.8.\(^6\) This system enables the construction of a biomechanical model and the assessment of the spatial-temporal variables of gait, as well as angular variation. Angular displacements of each joint were obtained according to the angle sequence proposed by Cardan.\(^39\) The reference or orthostatic position was considered the neutral position.

To eliminate the noise caused by the movement of markers, a low-pass Butterworth filter was used at a cutoff frequency of 6Hz.\(^40\)

Hip, knee, and ankle angular displacements were represented in percentage over the course of the gait cycle (0%–100%). To determine the start and end of the cycle, 2 consecutive IC events of the paretic foot were needed. This was achieved by observing the markers placed on the calcaneus or head of the 5th metatarsus. Raising the foot off the treadmill was determined by the marker placed on the head of the 5th metatarsus.

The events were defined based on the graphic representation of these markers on the y axis.\(^41\) These definitions were also determined for the nonparetic foot to provide data for the analysis of gait-related variables.

Only the 5 best cycles were selected for analysis. The spatial and temporal gait variables investigated were: velocity (m/s), cadence (steps/min), step length (m), cycle time (s), step time of the paretic and nonparetic leg (s), support time and swing time of the paretic leg (s), and interlimb symmetry ratio. To calculate the symmetry ratio, the following formula was used:\(^42\):

\[
\text{Interlimb symmetry ratio} = \frac{2\times (NP \text{ step time})}{NP \text{ step time} + P \text{ step time}}
\]

where NP stands for nonparetic and P stands for paretic.

In relation to angular variables, we investigated the angular displacements and amplitude of movement (AOM), in degrees, of the hip, knee, and ankle in the sagittal plane. AOM was obtained by subtracting the minimum from the maximum value reached. Hip and knee joint angles were analyzed in the following events: IC, maximum stance extension, and maximum swing flexion. For the ankle, the events were IC, maximum stance, and swing dorsiflexion and maximum plantar flexion.\(^11,14,42\)

Statistical Analysis

Data analysis was conducted using Statistica 7.0\(^d\) at a 95% confidence interval.\(^AQ:5\) Descriptive analysis was performed using central tendency and SD. The Shapiro-Wilk test was used to test normality. The parametric analysis of variance test with a randomized block design was also used. The Tukey post hoc test was used to detect statistically different measures. When the data did not
Table 1: Clinical and Demographic Characteristics of the Individuals (N=18)

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>Mean ± SD</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (y)</td>
<td>55.33±9.37</td>
<td>39–76</td>
</tr>
<tr>
<td>Body mass (kg)</td>
<td>70.39±10.16</td>
<td>50–91.5</td>
</tr>
<tr>
<td>Height (m)</td>
<td>1.61±0.09</td>
<td>1.47–1.76</td>
</tr>
<tr>
<td>Lesion time (mo)</td>
<td>34.00±22.80</td>
<td>8–84</td>
</tr>
<tr>
<td>Speed (m/s)</td>
<td>0.71±0.28</td>
<td>0.22–1.14</td>
</tr>
<tr>
<td>FIM</td>
<td>79.50±5.89</td>
<td>70–88</td>
</tr>
<tr>
<td>BBS</td>
<td>47.11±8.69</td>
<td>22–54</td>
</tr>
</tbody>
</table>

Abbreviation: BBS, Berg Balance Scale.

show normative distribution, the Friedman and Wilcoxon tests were used. The variables that showed normative distribution were cycle time, stride time in the paretic and nonparetic limb, stance time, intralimb symmetry, and maximum ankle dorsiflexion in the swing phase. The other variables analyzed did not exhibit normative distribution.

RESULTS

Out of a total of 23 individuals, 18 took part in this study (10 men and 8 women); 8 with left hemiparesis and 10 with right hemiparesis. Five subjects were excluded owing to alterations in blood pressure, heart rate, and fear of falling off the treadmill. The clinical and demographic characteristics were described (table 1).

Spatial-Temporal Variables

There was a decrease in cadence between 0% and 10% (91.12±18.29 vs 88.14±18.68) and between 5% and 10% (92.13±20.94 vs 88.14±18.68), but no significant differences were found (P=.160). Stance time increased with inclination and the differences were seen using the Wilcoxon test, comparing 0% and 5% (0.86±0.26 vs 0.81±0.20; P=.011) and 0% and 10% (0.86±0.26 vs 0.88±0.23; P=.021). Step length (P=.230), cycle time (P=.153), step time of the paretic (P=.150) and nonparetic limb (P=.330), paretic limb swing time (P=.118), and the interlimb symmetry ratio (P=.219) were minimally affected and showed no significant difference (table 2).

Angular Variables

The graphic representation of the hip, knee, and ankle joint angles in the sagittal plane during the gait cycle of the individuals assessed was made for each of the 3 treadmill conditions (fig 1).

Analysis of the hip joint, using Tukey ad hoc test, showed significant differences at IC, maximum extension, and maximum swing flexion. With an increase in inclination, there was an increase in flexion at IC at 0%, 5%, and 10% (P<.001) and

<table>
<thead>
<tr>
<th>Spatial-Temporal Variables</th>
<th>0%</th>
<th>5%</th>
<th>10%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cadence (steps/min)</td>
<td>91.12±18.29</td>
<td>92.13±20.94</td>
<td>88.14±18.68</td>
</tr>
<tr>
<td>Stride length (m)</td>
<td>0.84±0.26</td>
<td>0.86±0.26</td>
<td>0.85±0.25</td>
</tr>
<tr>
<td>Cycle time (s)</td>
<td>1.28±0.23</td>
<td>1.28±0.23</td>
<td>1.29±0.26</td>
</tr>
<tr>
<td>P step time (s)</td>
<td>0.69±0.16</td>
<td>0.69±0.16</td>
<td>0.71±0.17</td>
</tr>
<tr>
<td>NP step time (s)</td>
<td>0.58±0.10</td>
<td>0.58±0.10</td>
<td>0.58±0.11</td>
</tr>
<tr>
<td>Stance time (s)</td>
<td>0.83±0.21*</td>
<td>0.87±0.20*</td>
<td>0.88±0.23'</td>
</tr>
<tr>
<td>Swing time (s)</td>
<td>0.40±0.07</td>
<td>0.41±0.07</td>
<td>0.42±0.07</td>
</tr>
<tr>
<td>Interlimb symmetry ratio</td>
<td>0.91±0.09</td>
<td>0.92±0.09</td>
<td>0.90±0.09</td>
</tr>
</tbody>
</table>

NOTE: Values are presented as mean ± SD.
Abbreviations: NP, nonparetic lower limb; P, paretic lower limb.

*P<.011; †P<.021.
in swing at 0%, 5%, and 10% (P < .001). There was a decrease in maximum extension during the terminal stance between 0% and 5% (P = .021), 0% and 10% (P < .001), and 5% and 10% (P = .031). An increase was also observed in AOM between 0% and 10% (P < .001) and 5% and 10% (P = .002) (table 3).

In the knee, no differences were found in the maximum stance extension and the maximum swing flexion. According to the Tukey test, at IC, differences occurred between 0% and 10% and between 5% and 10% (both with P < .001). At maximum stance flexion, the differences occurred between 0% and 10% (P < .001) and between 5% and 10% (P = .004). AOM was different only between 0% and 10% (P = .040) (see table 3).

In the ankle, no significant maximum dorsiflexion differences were observed during stance or maximum plantar flexion. At IC there were differences only between 0% and 10% (Tukey ad hoc test P = .021). Maximum swing dorsiflexion only showed a difference between 5% and 10% (Wilcoxon test P = .040). No differences were found in AOM (see table 3).

DISCUSSION

Spatial-Temporal Variables

The spatial-temporal gait variables showed that higher degrees of inclination resulted in an increased stance time, whereas the other variables showed no significant alterations.

Studies that compared the overground and treadmill gait of hemiparetic individuals showed that the treadmill promoted an increase in stance time, likely owing to better body alignment resulting from the partial body-weight support system. In this study, the harness support was used only for safety purposes; however, the fact that the subjects held onto the front bar while walking may have interfered in body alignment. Furthermore, it was observed that stance time increased linearly as a function of the inclination percentage. Thus, it is believed that inclination was an additional factor that contributed to the increase of this variable and that the absence of alterations in the remaining variables may be explained by the constant speed and speed variability of individuals who walked at a comfortable, self-selected speed (0.22-1.14m/s).

Werner et al. assessed individuals with hemiparesis and observed an increase in stance time, a decrease in swing time, an increase in stride length, and decreased cadence. According to Visintin, Hesse, and colleagues, the use of partial body weight support on the treadmill produces an improvement in spatial-temporal gait variables such as speed, cadence, cycle length, and symmetry. However, in the present study, speed was maintained constant and body weight was not supported, a fact that may have interfered in cadence and in the other variables analyzed.

Angular Variables

In relation to the angular variables, it was observed that at IC and in the swing phase that there was an increase in hip and knee flexion and in ankle dorsiflexion only during IC. In healthy individuals, according to Leroux, Lay, and colleagues, higher degrees of inclination caused an increase in hip and knee flexion and in ankle dorsiflexion during IC and in the swing phase. Prentice et al. also observed an increase in flexion in the same joints during the swing phase. These authors explained that the aforementioned characteristics result from the need to elevate the lower limb and thrust the body forward while ascending. Similar results were found by Leroux et al. in a study investigating the effects of treadmill inclination on individuals with sequelae from spinal cord lesion in the hip joint. The other joints, however, showed no alterations, possibly owing to the limitations in knee and ankle joint adaptation caused by neural or biomechanic factors.

Leroux et al. also observed that in healthy individuals, the increase in hip flexion was used as adaptation to treadmill inclination, resulting in an increased stride length. In our study, the increase in hip flexion was not accompanied by an increase in stride length, but the increased flexion suggests that individuals with hemiparesis also use adaptation strategies to change limb trajectory and ensure IC on a more elevated surface. Stride length did not increase with inclination, likely because of the reduced hip extension. The decrease in hip extension and increase in flexion found in our study also may be explained by the need to elevate the limb to ensure contact at a higher level, as observed by Lay et al. It is important to point out that even though there was a reduction in extension, hip joint amplitude increased between 0% and 10% and between 5% and 10%. Given that individuals

Table 3: Angular Variables of the Hip, Knee, and Ankle During Treadmill Walking at Inclinations of 0%, 5%, and 10%

<table>
<thead>
<tr>
<th>Angular Variables</th>
<th>0%</th>
<th>5%</th>
<th>10%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>IC</td>
<td>23.11 ± 4.33*</td>
<td>26.12 ± 4.86*</td>
<td>29.99 ± 3.97*</td>
</tr>
<tr>
<td>Maximum stance extension</td>
<td>-12.41 ± 5.60*</td>
<td>-10.79 ± 6.14*</td>
<td>-9.28 ± 5.36*</td>
</tr>
<tr>
<td>Maximum stance flexion</td>
<td>25.42 ± 4.76*</td>
<td>28.02 ± 4.94*</td>
<td>31.84 ± 4.82*</td>
</tr>
<tr>
<td>AOM</td>
<td>37.83 ± 5.23*</td>
<td>38.80 ± 5.96*</td>
<td>41.12 ± 6.53*</td>
</tr>
<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>IC</td>
<td>17.59 ± 8.68*</td>
<td>18.52 ± 8.31†</td>
<td>22.16 ± 7.82‡</td>
</tr>
<tr>
<td>Maximum stance flexion</td>
<td>17.36 ± 11.14*</td>
<td>18.01 ± 10.72*</td>
<td>21.08 ± 11.24*</td>
</tr>
<tr>
<td>Maximum stance extension</td>
<td>3.20 ± 9.87</td>
<td>3.10 ± 9.36</td>
<td>2.79 ± 9.75</td>
</tr>
<tr>
<td>Maximum swing flexion</td>
<td>50.71 ± 13.65</td>
<td>51.53 ± 12.40</td>
<td>53.09 ± 10.53</td>
</tr>
<tr>
<td>AOM</td>
<td>47.51 ± 15.07‡</td>
<td>48.43 ± 14.24</td>
<td>50.30 ± 12.82‡</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>IC</td>
<td>-1.83 ± 8.08*</td>
<td>-1.51 ± 8.11</td>
<td>-0.59 ± 7.29‡</td>
</tr>
<tr>
<td>Maximum stance dorsiflexion</td>
<td>5.80 ± 5.11</td>
<td>6.13 ± 5.40</td>
<td>6.63 ± 5.38</td>
</tr>
<tr>
<td>Maximum plantar flexion</td>
<td>-5.41 ± 8.57</td>
<td>-6.20 ± 8.17</td>
<td>-4.66 ± 8.48</td>
</tr>
<tr>
<td>Maximum swing dorsiflexion</td>
<td>1.45 ± 8.08</td>
<td>1.74 ± 7.95‡</td>
<td>2.50 ± 7.42‡</td>
</tr>
<tr>
<td>AOM</td>
<td>11.21 ± 7.90</td>
<td>12.33 ± 7.77</td>
<td>11.29 ± 7.82</td>
</tr>
</tbody>
</table>

Note: Values are presented as mean ± SD.

*P < .001; †P = .021; ‡P = .031; ¶P = .002; ΔP < .001; §P = .040.
with hemiparesis show a decrease in hip flexion on IC and in swing and a consequent reduced amplitude, the increase in these parameters, which occurred at inclinations of 5% and 10%, may be a positive factor in promoting and stimulating a near-normative amplitude.5 Improving these parameters is the desired goal in the rehabilitation of hemiparetic individuals, given that such an outcome increases their ability to move body mass forward, thereby enhancing gait quality.

The knee joint showed that the individuals made IC with excessive flexion (17.59 ± 8.58 at 0%, 18.52 ± 8.31 at 5%, and 22.16 ± 7.82 at 10%), possibly owing to ischiobial or plantar flexor muscle spasticity or knee extensor weakness, which leads to flexion posture in the swing phase as well.4,8,9,10 There was no significant difference in maximum extension or maximum swing flexion, but inclination promoted a beneficial effect, given the significant increase in amplitude between 0% and 10%.

Qualitative analysis of IC showed that 10 of the 18 individuals made contact with the ankle plantar flexed, but with an increase in inclination to 10%, there was a significant decrease. In other words, there was a tendency toward the neutral position, which approaches the normative pattern seen in healthy individuals.4,6,8 According to Leroux et al,11 the increase in dorsiflexion occurs to accommodate the vertical orientation of the torso and the pelvis during the ascent phase, and this also may have occurred in hemiparetic individuals. This factor is positive, because it may contribute to improving dorsiflexion activation or mechanical properties (decreased stiffness) of muscles involved in plantar flexion and might also promote a decrease in the need to perform hip circumduction during the swing phase.

It is known that inclined surfaces pose significant challenges to the locomotor control system, in addition to resulting in substantial complications for subjects with pathologic gait caused by neurologic impairment. Thus, kinematic alterations and possible adjustments in limb trajectory may lead to a decrease in compensatory strategies and, in turn, to a decrease in energetic cost, greater access to certain environments, and reduced risk of falling.

Study Limitations

The small number of cameras used in this study precluded kinematic analysis of the hip and the nonparetic limb and small number of subjects. Additionally, no electromyographic assessment was performed, which could indicate the strategies used in the angular alterations occurred at 5% and 10% inclination.

CONCLUSIONS

According to these findings, we conclude that treadmill inclination promotes alterations in the angular variables of gait such as an increase in hip, knee, and ankle angle during IC and the swing phase, as well as an increase in AOM of the hip and knee. With respect to spatial-temporal variables, we conclude that inclination promotes minimal alterations, because only the stance time of the paretic lower limb was altered, evidenced by the increase of this variable. These results support further studies on the use of inclined treadmill walking in stroke rehabilitation and whether inclined walking training is effective in improving any of the functional limitations of patients poststroke.

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References


