The coordination of upper and lower limb movements during gait in healthy and stroke individuals

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1. Introduction

Human walking involves coordinated movements of all four limbs. The benefits of incorporating arm movements in gait rehabilitation are not known and difficult to investigate in patient populations with poor balance and reduced walking capacity. This study assessed the effect of supported (SUP) versus unsupported (UNSUP) arm movements on the coordination patterns present during walking in individuals with and without a stroke. Ten high functioning stroke subjects and 10 healthy subjects walked on a treadmill while swinging their arms naturally, and while holding onto handles that were either fixed in place or allowed to slide along horizontal handrails. Full-body kinematics were recorded, and arm–leg coordination was quantified using relative phase index, mean relative phase, and cross-correlation of hip and shoulder angle time series. No differences were observed in any measures of coordination between healthy and stroke subjects, indicating that the ability to coordinate arm and leg movements during walking remains preserved in high functioning stroke individuals. Coordination patterns were also unaffected by the use of sliding handrails, suggesting that this paradigm may be a suitable surrogate for natural arm movements if individuals are unable to walk without an external support. Stroke subjects were able to perform arm movements at a faster walking speed when using the handles than they were able to achieve without the handles, indicating that this paradigm may be useful in encouraging arm movements during gait rehabilitation.

ARTICLE INFO

Article history:
Received 24 January 2008
Received in revised form 20 May 2008
Accepted 22 May 2008

Keywords:
Locomotion
Walking
Hemiplegia
Arm swing

ABSTRACT

Human walking involves coordinated movements of all four limbs. The benefits of incorporating arm movements in gait rehabilitation are not known and difficult to investigate in patient populations with poor balance and reduced walking capacity. This study assessed the effect of supported (SUP) versus unsupported (UNSUP) arm movements on the coordination patterns present during walking in individuals with and without a stroke. Ten high functioning stroke subjects and 10 healthy subjects walked on a treadmill while swinging their arms naturally, and while holding onto handles that were either fixed in place or allowed to slide along horizontal handrails. Full-body kinematics were recorded, and arm–leg coordination was quantified using relative phase index, mean relative phase, and cross-correlation of hip and shoulder angle time series. No differences were observed in any measures of coordination between healthy and stroke subjects, indicating that the ability to coordinate arm and leg movements during walking remains preserved in high functioning stroke individuals. Coordination patterns were also unaffected by the use of sliding handrails, suggesting that this paradigm may be a suitable surrogate for natural arm movements if individuals are unable to walk without an external support. Stroke subjects were able to perform arm movements at a faster walking speed when using the handles than they were able to achieve without the handles, indicating that this paradigm may be useful in encouraging arm movements during gait rehabilitation.

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Subjects. All data presented are recorded at the matched speed, to enable comparisons between the populations.

Stage one is characterised by flaccid paralysis and stage six by the presence of near normal coordination patterns, yet still with some faulty patterns and timing with rapid and complex actions.

2. Methods

2.1. Subjects

Ten individuals who had sustained a stroke and 10 healthy adults participated in this study (Table 1). The stroke subjects had suffered only one stroke, had an incomplete motor recovery of the lower limb (Chedoke-McMaster Impairment Inventory score <7/7 for the leg and foot) and sufficient ability to use the sliding handles (Chedoke-McMaster Impairment Inventory score ≥3/7 for the arm and hand). The Chedoke-McMaster stroke assessment is a valid and reliable tool that quantifies the level of motor recovery [14]. Stage one is characterised by flaccid paralysis and stage six by the presence of near normal coordination patterns, yet still with some faulty patterns and timing with rapid and complex actions.

2.2. Equipment

The treadmill used was fitted with sliding handles mounted with a low-friction bearing system on horizontal bars (Fig. 1). Each handle contained a sensor that was sensitive to vertical forces. Subjects walked on the treadmill under three conditions: (1) holding onto handles that were fixed in place; (2) holding onto handles that were free to slide; (3) not holding onto the handles and having the arms hang freely by the sides of the body. In condition 1 arm movements were prevented resulting in ‘no arm movements’. In the other two conditions arm movements were encouraged, resulting in “supported arm movements” in condition 2 and “unsupported arm movements” in condition 3. The height and the medio-lateral location of the handrails were adjusted for each subject so that the position of the arms approximated that when they hung freely by the side of the body. All subjects wore a safety harness that was suspended from the ceiling and bore no weight but would provide support in the event of a fall.

2.3. Experimental procedure

The self-selected comfortable gait speed (CGS) of all subjects was measured using the self-paced capability of the treadmill, which allowed the user to modify walking speed at will [15]. The instructions were to walk at a comfortable speed, “as if walking around the grocery store”, and speed was averaged over 15–30 s to obtain CGS. The CGS of stroke subjects was slower than that of healthy subjects (Table 1). The self-selected fast gait speed (FGS) of all stroke subjects was also measured, where the instructions were to walk at a fast but safe speed, “as if rushing to catch the bus”.

Stroke subjects performed trials at CGS and FGS, and were only required to perform unsupported arm movement trials at FGS if they felt comfortable and safe doing so. Only one subject was able to walk with unsupported arm movement at FGS. Healthy subjects performed all three arm movement conditions at CGS and at a pre-determined slow speed, chosen to match the comfortable speed (MGS) of the stroke subjects (Table 1). The coordination of healthy subjects did not differ between MGS and CGS, and only the data from MGS is presented. A minimum of 12 strides was performed in every trial, and each condition was performed twice. The first and last strides of each trial were excluded, yielding a total of 20 strides per condition for the analysis. Three-dimensional full-body kinematic data were recorded at 120 Hz with the Vicon-512 system (Vicon Peak, Oxford, UK) using 40 reflective markers located on body landmarks as specified in the Plug-In-Gait model from Vicon. The vertical forces exerted on the handles with no and supported arm movements were sampled at 1080 Hz.

2.4. Data analysis

Marker trajectories were labelled offline to create a link-segment model of the body. Joint angles were computed with the Plug-In-Gait model. Marker positions and joint angles were low pass filtered at 10 Hz. Stride cycles were identified from Vicon. The vertical forces exerted on the handles with no and supported arm movements were sampled at 1080 Hz.

Table 1

Individual subject characteristics for all stroke patients, plus averages for stroke and healthy groups

<table>
<thead>
<tr>
<th>Sex</th>
<th>Age (years)</th>
<th>Height (m)</th>
<th>Weight (kg)</th>
<th>Comfortable gait speed (m s⁻¹)</th>
<th>Fast gait speed (m s⁻¹)</th>
<th>Time since stroke (weeks)</th>
<th>Affected leg/foot</th>
<th>Chedoke-McMaster score (max = 7)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stroke (N = 10)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>S1</td>
<td>F</td>
<td>67</td>
<td>1.55</td>
<td>93</td>
<td>0.38</td>
<td>0.55</td>
<td>23</td>
<td>Left</td>
</tr>
<tr>
<td>S2</td>
<td>M</td>
<td>63</td>
<td>1.76</td>
<td>88</td>
<td>0.52</td>
<td>0.80</td>
<td>86</td>
<td>Left</td>
</tr>
<tr>
<td>S3</td>
<td>M</td>
<td>62</td>
<td>1.83</td>
<td>104</td>
<td>0.71</td>
<td>1.10</td>
<td>89</td>
<td>Right</td>
</tr>
<tr>
<td>S4</td>
<td>M</td>
<td>72</td>
<td>1.69</td>
<td>72</td>
<td>0.85</td>
<td>1.20</td>
<td>23</td>
<td>Left</td>
</tr>
<tr>
<td>S5</td>
<td>M</td>
<td>66</td>
<td>1.72</td>
<td>77</td>
<td>0.85</td>
<td>1.12</td>
<td>28</td>
<td>Right</td>
</tr>
<tr>
<td>S6</td>
<td>M</td>
<td>73</td>
<td>1.75</td>
<td>87</td>
<td>0.75</td>
<td>0.92</td>
<td>56</td>
<td>Left</td>
</tr>
<tr>
<td>S7</td>
<td>M</td>
<td>45</td>
<td>1.85</td>
<td>95</td>
<td>0.62</td>
<td>0.75</td>
<td>108</td>
<td>Right</td>
</tr>
<tr>
<td>S8</td>
<td>M</td>
<td>49</td>
<td>1.75</td>
<td>106</td>
<td>1.15</td>
<td>1.3</td>
<td>70</td>
<td>Right</td>
</tr>
<tr>
<td>S9</td>
<td>M</td>
<td>67</td>
<td>1.76</td>
<td>73</td>
<td>0.34</td>
<td>0.61</td>
<td>26</td>
<td>Left</td>
</tr>
<tr>
<td>S10</td>
<td>M</td>
<td>54</td>
<td>1.73</td>
<td>76</td>
<td>0.61</td>
<td>1.00</td>
<td>51</td>
<td>Right</td>
</tr>
<tr>
<td>Mean ± S.D.</td>
<td>–</td>
<td>62 ± 10</td>
<td>1.74 ± 0.08</td>
<td>87 ± 12</td>
<td>0.68 ± 0.24</td>
<td>0.94 ± 0.25</td>
<td>56 ± 31</td>
<td>–</td>
</tr>
</tbody>
</table>

Healthy (N = 10)

Mean ± S.D. – 62 ± 6 1.69 ± 0.11 74 ± 10 0.70 ± 0.23

Note that Chedoke-McMaster Impairment Inventory scores were not assessed for subject S1.

* This is slower than the self-selected comfortable gait speed of the healthy subjects, which was 1.19 ± 0.26 m s⁻¹. Slower speeds were selected to match those of the stroke subjects. All data presented are recorded at the matched speed, to enable comparisons between the populations.
each stride and averaged over all strides for each subject. Arm movement cycles were identified from local maxima in the sagittal plane shoulder angle time series, with arm cycle duration and arm movement amplitude determined in an identical fashion.

For each trial, a cross-covariance analysis was performed on contralateral hip and shoulder joint angle time series to assess the correlation between the two signals. The strength of the peak correlation and the time shift (lag) at which this occurred were determined. A negative lag indicated that maximal flexion of the shoulder occurred before maximal flexion of the hip.

Additional measures of coordination included the point estimate of relative phase and the frequency relation between ipsilateral arm and leg movements [5,6]. The point estimate of relative phase was calculated for each stride as the relative occurrence of peak shoulder angle, and averaged over all strides to obtain the mean relative phase (MRP). In the case that there was more than one peak in shoulder angle for a given stride cycle then the first peak was used [5,6]. The frequency relation was obtained from the power spectral density (PSD) function of the hip and shoulder angle time series. Signals in the PSD of less than 0.2 Hz were removed [16] and the PSD was divided by the mean power over the 0.2–2.5 Hz frequency range. Consistent with previous results [1,3] the frequency with the largest power in the hip angle time series corresponded to stride frequency ($P_1$) and step ($P_2$) frequencies was determined, and used to calculate relative power index (RPI):

$$\text{RPI} = \frac{P_1 - P_2}{P_1 + P_2}$$

RPI can range from −1 (representing a locking of arm movement frequency onto step frequency) to +1 (representing a locking of the arms onto stride frequency). For each trial, a cross-covariance analysis was performed on contralateral hip and contralateral shoulder angles for trials performed with supported (SUP) and unsupported (UNSUP) arm movements. Zero lag is indicated by a vertical dashed line in B. The boundaries of the boxes indicate 25th and 75th percentile, and the line within the box is the median value. Whiskers indicate the 10th and 90th percentiles, and outlying points are indicated as black dots.

2.5. Statistical analysis

2.5.1. Within group analyses

A two-way repeated measure analysis of variance (ANOVA) of Side X Condition was performed for each group of subjects. Additional repeated measures ANOVAs were used to compare CGS and FGS in the stroke subjects. Unsupported arm movements were not performed without the use of the handrails at FGS, therefore this condition was not compared across speeds.

2.5.2. Between group analyses

There were no differences in any variables between the left and right side of the healthy subjects. Therefore, the data from the left side of healthy subjects was used for comparisons across groups. A two-way repeated measure ANOVA of Group X Condition was performed once to compare the paretic side of stroke subjects to the healthy subjects, and again to compare the non-paretic side of stroke subjects to the healthy subjects. This comparison was performed with data recorded at CGS in stroke and MGS in healthy subjects.

2.5.3. Parametric assumptions

Normality was assessed using the Shapiro–Wilks test and visual inspection of quantile–quantile plots. Arm movement amplitude and cycle duration, stride amplitude and duration, and MRP were all normally distributed and analysis proceeded as described above. Cross-covariance correlation and lag, and RPI were non-normally distributed and were assessed using a generalised form of non-parametric statistics, as detailed by Thomas et al. [17]. When the ANOVA yielded a significant interaction post hoc analyses were completed using $t$-tests with Bonferroni correction for multiple comparisons. Where there was no significant interaction the main effects for each factor are reported. Statistical analyses were done in SPSS and Statistica with a level of statistical significance set at $p < 0.05$.

3. Results

3.1. Stroke versus healthy subjects

The vertical forces exerted on the handles with no and supported arm movements were low, and did not significantly differ between stroke and healthy subjects. On average, healthy subjects exerted 2.7 ± 1.5% body weight (BW), and the force did not change across the two arm movement conditions ($p = 0.147$). Stroke subjects exerted less force on the handrails with supported than with no arm movements ($p = 0.029$), but this difference was small in magnitude, decreasing from 4.9 ± 2.9 to 4.1 ± 3.0% BW on the paretic side, and from 4.9 ± 2.9 to 3.6 ± 3.0% BW on the non-paretic side.

Both healthy and stroke subjects took shorter and more frequent strides with unsupported arm movements, and swung their arms more frequently. Arm cycle duration correlated very well with stride duration in all conditions ($r = 0.933$ for stroke and $r = 0.816$ for healthy subjects, $p < 0.001$). The pattern and amplitude of shoulder movement was similar for the two arm movement conditions. The strength of the correlation between hip and contralateral shoulder angles ranged from 0.17 to 0.96 for...
stroke (mean = 0.72 ± 0.25) and from 0.21 to 0.97 for healthy subjects (mean = 0.75 ± 0.23; Fig. 2A). The correlations displayed considerable negative skew for each condition due to a few subjects with extreme low values. These less coordinated individuals were evenly distributed between stroke and healthy subjects. The time lag at which the peak correlation occurred was close to zero for all conditions (Fig. 2B), indicating an in-phase coordination pattern between contralateral arm and leg. There was no difference between stroke and healthy subjects in either the magnitude or the lag of the cross-covariance. These measures were also similar across all arm movement conditions and across both limb pairs. These results are supported by an MRP close to 180° for all conditions, indicating out-of-phase coordination between ipsilateral limbs, and an RPI closer to 1 than 0. Neither of these measures of coordination differed between groups of subjects nor across sides of the body or arm movement conditions. All four coordination variables (cross-covariance correlation and lag, MRP and RPI) showed weak and generally insignificant relationships with both walking speed and the amplitude of arm movement (Table 2).

### 3.2. Effect of speed

Stroke subjects exerted more force on the handles at FGS (5.3 ± 3.0% BW) than at CGS (4.7 ± 2.7% BW; p = 0.015), but the mean difference was small. Comparisons between the two speeds in stroke subjects do not include walking with unsupported arm movements, as these trials were not performed at FGS. Strides were shorter (p = 0.001) and more frequent (p = 0.024) at FGS but remained similar across the two conditions. The frequency of arm movements increased (p = 0.003) at FGS, while the amplitude increased only

![Figure 3: Effects of speed: arm movement duration (A) and amplitude (B), and mean relative phase (C) and relative phase index (D) between ipsilateral limbs during walking with supported arm movements at comfortable gait speeds (CGS) and fast gait speeds (FGS). Data are shown for the paretic (P) and non-paretic (NP) sides of the body. Error bars represent ±S.D. Significant (p < 0.05) differences between speeds are indicated with stars (*)�].
for the non-paretic arm ($p = 0.017$, Fig. 3A and B). The amplitude of the arm movements was greater for the non-paretic arm than the paretic arm at both CGS ($p = 0.030$) and FGS ($p = 0.001$, Fig. 3B). No measures of arm–leg coordination changed at the faster speed; the magnitude and lag of the cross-covariance correlation were similar across speeds, as were MRP and RPI (Fig. 3C and D).

4. Discussion

Arm–leg coordination during walking was unaffected by the use of sliding handles, supporting our first hypothesis and suggesting that this paradigm may be suitable for use in rehabilitation. Using sliding handles allowed stroke subjects to perform arm movements while walking at a faster speed than they were able to achieve without an external support. In the current study nine out of 10 stroke subjects could not walk at the faster speed if external support was not provided, but with the sliding handles they were able to walk safely and perform arm movements. This allowed us to assess the degree of arm–leg coordination in stroke subjects at speeds faster than those previously examined in patient populations. The coordination was similar at the faster speed. This did not support our second hypothesis that the coordination patterns of stroke patients would improve with increased speed. The lack of improvement was likely due to the already high level of coordination present at the slower speed.

Although arm–leg coordination during walking has been examined in subjects with stroke [5–7] it has never been directly compared with the coordination observed in healthy individuals. The ability to coordinate the arms and legs during walking was unaffected by stroke in the individuals in this study. The degree of coordination varied across individuals, however this was the case for both stroke and healthy subjects and there were no differences between the two groups. There was no strong relation between the degree of coordination and either walking speed or the amplitude of arm movements, and no other distinctive features of the less coordinated individuals could be identified. When performing isolated single-limb voluntary movements of the forearm and lower leg stroke subjects were less coordinated than healthy subjects [18] however this task has been shown to involve activity over a widely distributed network of brain regions [19]. The ability of stroke subjects to coordinate homologous limb pairs in the present study may reflect a decreased reliance on cortical input in the more automatized task of walking. The absence of differences between groups does not preclude that more acute or more severely affected stroke subjects would demonstrate coordination deficits. One limitation of the present study is that the stroke subjects studied were all relatively high functioning, with an average gait speed of 0.68 ± 0.24 m s$^{-1}$ and Chedoke-McMaster scores of at least 4/6 for the leg, foot, arm and hand (Table 1). The paradigm of sliding handrails may prove useful in assessing movement patterns in more impaired populations who cannot perform unsupported arm movements during walking.

The stroke subjects studied were able to coordinate the paretic and non-paretic sides of their body equally well. This agrees with one previous study [5] but contrasts others which have reported decreased coordination on the paretic side [26–31]. Heterogeneity in gait disorders is well documented for the stroke population, however the subjects studied by Ford et al. [6] do not appear to be lower functioning than the subjects assessed here. It is worth noting that Ford et al. [6] instructed subjects to move their arms and legs in time with an auditory beat, and this may have altered the task demands sufficiently to result in the observed difference. Detailed information on neither the subjects nor the conditions under which coordination was studied were provided by Wagenaar and van Emmerik [7], therefore the discrepant results may reflect either of these factors.

Stroke subjects moved their non-paretic arm more than their paretic arm, supporting the findings of Ford et al. [5]. This may be a direct result of the stroke, however when one arm of healthy subjects was restrained during walking, the amplitude of the non-restrained arm also increased [4], suggesting that this may be an adaptive strategy aimed at maintaining coordination between upper and lower body. Interestingly, while the frequency of arm movements increased with speed, the amplitude of the movement in the paretic arm did not increase. This may suggest that the paretic arm is swinging at maximal amplitude when walking at comfortable speed and has no capacity to increase its movement amplitude. Dissociation between the amplitude and timing of arm movements has also previously been reported in healthy subjects, where movement amplitude decreased but movement frequency was maintained when a mass was added to one arm during walking [3].

5. Conclusions

Our results support previous studies which reported that high functioning stroke subjects are capable of performing arm movements during gait. We also directly compared the coordination of stroke subjects to that of healthy individuals under identical conditions and demonstrated that the degree of arm–leg coordination was not affected by stroke in this population. The arms and legs were similarly coordinated with and without the use of sliding handles, and stroke subjects could walk at a faster speed with the handrails than without. The sliding handles may therefore be useful in gait rehabilitation protocols, where it has been suggested that performing arm movements may be beneficial to the patients [10,11].

Acknowledgements

We would like to thank Eric Johnston and Christian Beaudouin for their help in developing the treadmill for this project, and Maxim Hanna and Rachel Kizony for their help with data collection. This study was funded by the Canadian Institute of Health Research and the Multidisciplinary Team in Locomotor Rehabilitation (CIHR Team grant, Regenerative and Nanomedicine). J. Stephenson was the recipient of studentships from the Multidisciplinary Team in Locomotor Rehabilitation, McGill Faculty of Medicine and the CIHR-MENTOR Program. A. Lamon-tagne is funded by CIHR and the Canadian Foundation for Innovation. Study sponsors had no role in the study design, in the collection, analysis and interpretation of data, in the writing of the manuscript, and in the decision to submit the manuscript for publication.

Conflict of interest

None.

References


