The effect of arm movements on the lower limb during gait after a stroke
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**Abstract**

The purpose of this study was to examine the influence of arm movements on lower limb movement and muscle activation during treadmill walking after a stroke. Ten high functioning stroke 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, along with bilateral surface electromyography from lower limb muscles. Arm movements influenced lower limb muscle activity but had little effect on movement patterns at the joints. When handles were present a small amount of weight was borne through the upper limbs, and for stroke subjects this was reduced when the handles were free to slide. Activity of proximal leg muscles during stance was affected by the weight borne through the upper limbs, increasing when arm movements were performed. Soleus activity during stance was greatest with unsupported arm movements. In stroke subjects, early stance tibialis anterior activity in the paretic leg was greatest with no arm movements, and early swing tibialis anterior activity in both legs was greatest with unsupported arm movements. Many of the changes in muscle activation appeared to be due to changes in postural stability that occurred when performing arm movements. Overall, results support further study of the long-term changes associated with the inclusion of arm movements in gait rehabilitation protocols.

**Keywords:** Walking, Arm swing, Upper limb, Muscle activation, Electromyography, Hemiparesis, Rehabilitation

**1. Introduction**

Human walking involves active movements of the upper limbs [1–3]. Constraining the arms influences the gait pattern in both healthy [4–6] and patient [7,8] populations, and there is indirect support for the existence of coupling between upper and lower limb muscular activation patterns [5,9]. Accordingly, upper limb involvement altered lower limb muscle activation during recumbent stepping movements in healthy individuals [10] and during passive reciprocal leg movement in individuals with spinal cord injury [11]. While there are anecdotal reports from spinal cord injured patients that stepping was facilitated [12] and patterns of lower limb muscle activity improved [13] when arm swing was performed during walking, this has not been systematically investigated.

One barrier to directly investigating the effects of arm movements on the lower limb during walking in patient populations is the ability of these individuals to walk safely without using their arms to hold on to fixed external devices for balance support. We fitted a treadmill with bilateral handles that could slide in the horizontal direction, allowing arm movements in the sagittal plane while still assisting with balance maintenance. When using the sliding handles, individuals who had sustained a stroke were able to incorporate arm movements into their gait pattern at faster speeds than they were otherwise able to do so [14]. This suggests that these handrails may have potential for use in gait rehabilitation, where speed is important [15], however a thorough understanding of their effect on the gait pattern is necessary before recommendations can be made.

The purpose of this study was to assess the influence of arm movements performed with and without the sliding handles on lower limb kinematic and muscle activation patterns during walking in individuals who have sustained a stroke. We hypothesized that performing arm movements during treadmill locomotion would result in stride characteristics and lower limb muscle activation in stroke patients that were more similar to those observed in healthy individuals.

**2. Methods**

**2.1. Subjects**

Ten individuals who had sustained a stroke (nine males; mean (SD) age 62 (10) years) participated in this study. Time since stroke ranged from 23 to 108 weeks. Detailed subject characteristics are provided in Stephenson et al. [14]. Data from 10 healthy adults (six males) of a similar age (62 (6) years) were used for comparison. The stroke subjects had incomplete motor recovery of the lower limb (Chedoke–McMaster...
Impairment Inventory [16] score < 7/7 for the leg and foot) and sufficient ability to use the handrails (Chedoke-McMaster > 5/7 for the arm and hand), and are the same subjects as those previously studied [14]. The project was approved by the institutional ethics committee and subjects provided free and informed consent prior to participation.

2.2. Equipment

The treadmill used was fitted with custom built sliding handles that were mounted with a low-friction bearing system on horizontal bars, allowing arm movements in the sagittal plane [14]. Handles were fitted with a load cell that was sensitive to force in the vertical direction (Omega, Stamford, CT). 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; and (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 mediolateral 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 suspended from the ceiling that bore no weight but would provide support in the event of a fall.

2.3. Experimental procedure

The comfortable gait speed (CGS) of all subjects was measured using a self-paced capability of the treadmill, which allowed the subjects to modify gait speed at will [17]. The mean (SD) CGS of stroke subjects (0.68 (0.24) m s\(^{-1}\)) was slower than that of healthy subjects (1.19 (0.26) m s\(^{-1}\)). For nine stroke subjects CGS was measured with both no arm movements and supported arm movements, and did not differ between the two conditions (0.72 m s\(^{-1}\) vs. 0.69 m s\(^{-1}\), \(p = 0.422\)). Treadmill speed was then controlled externally, as per a conventional treadmill, for all experimental trials. Stroke subjects performed trials at their pre-determined CGS. Healthy subjects performed trials at an assigned slow speed that matched the CGS of the stroke subjects (0.70 (0.23) m s\(^{-1}\)). A minimum of 10 strides was performed in every trial, and each condition was performed twice, resulting in 20 strides for analysis.

Kinematic data were sampled at 120 Hz with the Vicon-512 system (Vicon Peak, Oxford, UK), using forty reflective markers placed on specific anatomical landmarks (Vicon Plug-In-Gait). Muscle activation was recorded as surface electromyography (EMG) from the soleus, tibialis anterior, semitendinosus and quadriceps muscles on the left side of healthy subjects walking at a matched speed. Error bars represent +1SD. Significant (\(p < 0.05\)) differences between arm movement conditions are indicated with asterisks (*).

EMG signals were band-pass filtered (20–400 Hz), rectified, low-pass filtered (10 Hz) to generate a linear envelope, and normalized to 100 points per stride. Signal amplitude was normalized to the peak amplitude recorded with no arm movements. EMG data were quantified for each stride as the mean value of the normalized signal over pre-determined time-windows (Fig. 3) chosen to capture selected bursts of activity. These quantified values were averaged over all 20 strides within a given condition for each subject. Ensemble averaged EMG profiles were obtained in the same way as for kinematic data.

2.4. Data analysis

Joint angles were computed as the relative angle between adjacent segment vectors (Vicon Plug-In-Gait), and were low-pass filtered at 10 Hz using a dual-pass Butterworth filter (Matlab, MathWorks Inc., Natick, MA) for zero phase lag. Stride duration was defined as the time between two successive ipsilateral foot strikes.

The range of motion (RoM) at the hip, knee and ankle was calculated for each stride as the peak-to-peak change in the angular displacement at these joints in the sagittal plane. Joint angle time series data were normalized to 100 points per stride and averaged over all 20 strides within a condition to obtain ensemble averaged joint angle profiles for each subject. These profiles were then averaged over all subjects within each group.

RoM was lowest with unsupported arm movements for both the ankle and hip joints. Knee RoM did not differ with arm movement condition, and hip RoM was lowest with unsupported arm movements. The changes in ankle and hip RoM are likely to be related to the reduced stride length observed with unsupported arm movements.

2.5. Results

3.1. Lower limb kinematics

Both groups of subjects took more frequent and shorter strides when performing unsupported arm movements (Fig. 1). There was a significant main effect of arm movement condition for ankle RoM in stroke (\(p = 0.029\)) and healthy (\(p = 0.012\)) subjects (Fig. 2). Although no post hoc comparisons were significant the tendency was for reduced ankle RoM with unsupported arm movements. Knee RoM did not differ with arm movement condition, and hip RoM was lowest with unsupported arm movements for both groups of subjects. The changes in ankle and hip RoM are likely to be related to the reduced stride length observed with unsupported arm movements.

3.2. Lower limb muscle activity

Ensemble averaged EMG profiles are shown in Fig. 3. In stroke subjects, early stance quadriceps activity was larger with
supported than with supported arm movements ($p = 0.001$; Fig. 4A). This difference was not present in healthy subjects ($p = 0.184$). Semitendinosus activity over the same period was greatest with unsupported arm movements for both groups. However, while activation also increased from no arm movements to supported arm movements in stroke subjects ($p = 0.012$), this difference was not present in healthy subjects (Fig. 4C). Early stance tibialis anterior activity did not change with arm movement condition in healthy individuals nor in the paretic leg of stroke subjects, but was greater with no arm movements than with supported arm movements in the non-paretic limb of stroke subjects ($p = 0.007$; Fig. 4E). Early stance soleus activation was greater with supported arm movements than with no arm movements, and greater again with unsupported arm movements for both stroke and healthy subjects (Fig. 4G). In healthy subjects the soleus activation in the second half of stance was slightly greater with no arm movements than with supported arm movements ($p = 0.046$), but this change was not present in stroke subjects (Fig. 4H).

Quadriceps activity during pre-swing and initial swing was not affected by arm movement condition in healthy subjects ($p = 0.094$), but was greater with unsupported arm movements than with supported arm movements in both the paretic ($p = 0.025$) and non-paretic ($p < 0.001$) legs of stroke subjects. In the paretic leg, activity was also greater with unsupported arm movements than with no arm movements ($p < 0.001$, Fig. 4B). Semitendinosus activity at the end of swing was greatest with unsupported arm movements for both groups of subjects (Fig. 4D). Tibialis anterior activity in early swing was greatest with unsupported arm movements in both legs of stroke subjects but was not affected by arm movement condition in healthy subjects ($p = 0.612$; Fig. 4F).

3.3. Weight borne on handrails

The vertical force exerted on the handrails was not significantly different between stroke and healthy subjects, though it was more variable among stroke subjects (Fig. 5). Stroke subjects exerted more force on the handrails in the second half of the gait cycle when no arm movements were performed compared to when supported arm movements were performed (Fig. 5B). Although this difference was statistically significant it was small in magnitude, with changes in the order of magnitude of 2% BW. It should be noted that the power for these comparisons was lower than for other outcome variables, due to the lower number of subjects on which force data were collected.

4. Discussion

Performing arm movements during treadmill walking influenced stride characteristics and patterns of lower limb muscle activation, but did not influence the overall patterns of movement at the joints. The observed changes may be a result of the arm movements themselves, the changes in postural stability that occurred as a function of the level of external support, or a combination of both these factors.

Subjects took longer and less frequent strides when the arms were restrained through holding onto the handles. At slow speeds such as those adopted by our subjects, Eke-Okoro et al. [5] also reported that healthy subjects took longer and less frequent strides when the arms were restrained, however these authors restrained the arms by strapping them to the side of the body, and did not provide external support. This suggests that the effect may be more related to the presence or absence of arm movements per se, rather than to the provision of external support.

4.1. Muscle activity during stance

During stance, the bursts of activity in semitendinosus and quadriceps muscles tended to increase when unsupported arm movements were performed. Extensor muscle activity during stance is sensitive to the amount of loading on the body [19], and this result is likely due to the increased weight bearing through the
legs when the external support provided by the handrails was removed. Although the increased muscle activity may promote strength gains and may more realistically prepare a stroke patient for overground walking, the continuation of high levels of activity throughout stance may be detrimental to developing appropriate phasic muscle activity [20]. Interestingly, stroke subjects displayed an additional burst of semitendinosus activity in the non-paretic leg at the end of stance, the magnitude of which was increased when arm movements were performed. This activity does not appear to be related to enhanced hip extension that is sometimes [e.g. 21] reported on the non-paretic side. It is possible that the activity was a compensatory strategy aimed at stabilizing the pelvis and upper body on the supporting non-paretic leg while the contralateral paretic arm was swung forward.

In stroke subjects, soleus activation in early stance was lowest when walking with no arm movements, increased with supported arm movements and increased again with unsupported arm movements. Although soleus activity is sensitive to load [19], healthy subjects exhibited the same pattern of activation despite having no differences in weight bearing between walking with no arm movements and supported arm movements. This suggests that the change in soleus activation was not solely due to differences in weight bearing. This effect does not appear to be related to changes in joint angles at foot strike or the placement of the foot in relation to the body. The traditionally referenced pattern of activation for the soleus is more similar to that seen with no arm movements, where activity is minimal in early stance [22]. There are previous reports of sustained plantarflexor activity throughout stance during walking on a treadmill [23], particularly at slow speeds [24]. The fact that the additional activity occurs at a phase of the gait cycle where ankle plantarflexors are working eccentrically indicates that it likely represents an inefficient movement strategy. Perhaps treadmill walking constitutes a postural threat that is countered by increasing soleus activity throughout stance, and this effect is exacerbated when walking at slow speeds, or when the balance support and tactile cues usually provided by handrails are removed. This explanation is supported by the general increase in activity of all lower limb muscles that is evident across the stance phase in stroke patients (Fig. 3).

4.2. Muscle activity during swing

The most striking influence of arm movement condition on muscle activation during swing was on the tibialis anterior muscle in stroke subjects, where activity was increased during walking with unsupported arm movements. Insufficient activation of the tibialis anterior is often observed at this point of the gait cycle in stroke patients [25], and leads to an insufficient toe clearance for swing. These results suggest that preventing arm movements may exacerbate this problem for patients.

In stroke subjects, quadriceps activity at the beginning of swing increased when unsupported arm movements were performed. At this point in the gait cycle the hip is generating positive power though concentric activity in the hip flexors [20,26]. The increased activity

![Fig. 3. Ensemble averaged EMG profiles from quadriceps (QUAD), semitendinosus (ST), tibialis anterior (TA) and soleus (SOL) during walking with no arm movements (solid line), while performing supported arm movements using sliding handrails (long dashes) and while performing unsupported arm movements (short dashes). Data are shown for the paretic and non-paretic sides of stroke subjects walking at a comfortable speed, and for the left side of healthy subjects walking at a matched speed. EMG amplitude was normalized to peak amplitude in the fixed arm movement condition prior to ensemble averaging. Horizontal lines indicate the periods over which muscle activity was quantified.](image-url)
may represent increased rectus femoris activity during early swing, increasing the propulsive pull of the leg into swing. In both groups of subjects there was also increased activity in the semitendinosus at the end of swing with unsupported arm movements. The possibility that these results were due to a faster acceleration of the limb through swing was examined and ruled out, and the vertical excursion of the centre of mass was also similar across conditions.

Previous studies have shown effects of rhythmic arm movements on lower limb muscle activation [10,11]. It is possible that the changes in muscle activation observed in the current study were partly caused by altered transmission in propriospinal neural pathways when arm movements were performed [27,28]. However, patterns of muscle activity most resembling those typically reported for overground walking were generally observed in the no arm movement condition, suggesting that the postural instability induced by removing external support may have had a greater effect on the patterns of muscle activity observed.

4.3. Limitations

The forces applied to the treadmill handles were measured only in the vertical direction, and only in a subset of subjects. The
handles could have been used to help propel the body forward, but this remains to be confirmed with the measurement of shear forces. Due to the low number of subjects, a Type II error may have occurred in comparing the amount of the weight borne though the arms across arm movement conditions. Furthermore, the stroke subjects studied were relatively high functioning, and further studies are needed to verify if the present results can be extended to more severely affected patients.

5. Conclusions

Overall, the results of this study indicate that using sliding treadmill handrails resulted in several changes in muscle activation patterns, and support further investigation into the use of these handrails in gait rehabilitation. The effects of performing arm movements occurred concomitantly with the effects of changes in postural stability. The resulting changes in patterns of muscle activity may facilitate transfer to overground walking where external support is not readily available. Alternatively, it is possible that the increased challenge to postural stability may result in the development of an inefficient gait pattern. Further study is required to determine the long-term effects of the observed changes on gait rehabilitation outcomes.

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Conflict of interest statement
No authors have any conflicts of interest to declare.

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