Alterations in muscle activation patterns during robotic-assisted walking

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Abstract

Objective. The goal of this study was to compare the muscle activation patterns in various major leg muscles during treadmill ambulation with those exhibited during robotic-assisted walking.

Background. Robotic devices are now being integrated into neurorehabilitation programs with promising results. The influence of these devices on altering naturally occurring muscle activation patterns utilized during walking have not been quantified.

Methods. Muscle activity measured during 60 s of walking was broken up into individual stride cycles, averaged, and normalized. The stride cycle was then broken up into seven distinct phases and the integrated muscle activity during each phase was compared between treadmill and robotic-assisted walking using a multi-factor ANOVA.

Results. Significant differences in the spatial and temporal muscle activation patterns were observed across various portions of the gait cycle between treadmill and robotic-assisted walking. Activity in the quadriceps and hamstrings was significantly higher during the swing phase of Lokomat walking than treadmill walking, while activity in the ankle flexor and extensor muscles was reduced throughout most of the gait cycle in the Lokomat.

Conclusions. Walking within a robotic orthosis that limits the degrees of freedom of leg and pelvis movement leads to changes in naturally occurring muscle activation patterns.

Relevance

An understanding of how robotic-assisted walking alters muscle activation patterns is necessary clinically in order to establish baseline patterns against which subject/s with neurological disorders can be compared. Furthermore, this information will guide further developments in robotic devices targeting gait training.

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1. Introduction

Over the last decade, there has been growing support for the use of manual-assisted treadmill training in neurorehabilitation programs (Dobkin, 1999). Preliminary studies have found that individuals who receive body-weight supported treadmill training following stroke (Hesse et al., 1994, 1997) and spinal cord injury (Wernig and Muller, 1992; Wernig et al., 1999; Dietz et al., 1995) demonstrate improved electromyographic (EMG) activity during locomotion (Visintin et al., 1998; Dietz et al., 1995), walk more symmetrically (Hassid et al., 1997; Hesse et al., 1997), are able to bear more weight on their legs (Wernig and Muller, 1992) and experience higher returns in functional walking ability when compared to patients who receive conventional gait training (Wernig...
et al., 1999; Visintin et al., 1998). It is postulated that treadmill training is a therapeutic paradigm that effectively activates afferent receptors in the lower limbs, generating the necessary sensory feedback needed to train central pattern generators in the spinal cord believed to underlie locomotion (Forssberg, 1979; Grillner, 1979, 1981). The primary limitation with manual-assisted, body-weight supported treadmill therapy is that a training session relies on several physical therapists to manually assist the patient's leg movements through the gait cycle, a protocol that is not cost effective in the current health care system. Furthermore, training sessions tend to be short because of the physical demands on the therapists, which may limit the full potential of the treatment.

Recent advances in robotics attempt to combat this problem by automating gait training with actuated devices (Colombo et al., 2000; Hesse and Uhlenbrock, 2000). These systems, used in conjunction with a body-weight support system, help move the individual's legs through well-specified and consistent gait patterns as s/he walks on a treadmill (Colombo et al., 2000). The potential advantages of using robotic devices include patient safety, repeatability, unlimited duration of training, and hands-free operation by a single therapist. In addition, because the patient is secure in the device removing the potential for falls, s/he is more apt to adopt a natural gait pattern rather than a guarded, cautious gait in order to ensure stability and prevent falls.

One potential limitation with robotic-assisted gait training is that the degrees of freedom through which the subject is able to walk is necessarily limited due to safety and control complexity. As a result, individuals are unable to execute some of the key gait determinants (Saunders et al., 1953) which may alter the set of muscle activation patterns required for stable over-ground walking.

The objective of this study was to determine whether muscle activation patterns exhibited while walking within a robotic orthosis are different from those demonstrated during treadmill ambulation. Furthermore, we sought to establish a set of “normative” muscle activation patterns which could be used as baseline measures against which the muscle activation patterns exhibited by individuals with gait disorders could be compared to. Portions of this work have been previously reported in abstract form (Hidler et al., 2003).

2. Methods

2.1. Subjects

A total of seven healthy subjects (3 male, 4 female) with no known neurological injuries or gait disorders participated in the study (mean age: 26.8 years; range: 24–30). All experimental procedures were approved by the Research Review Board at the National Rehabilitation Hospital and the Institutional Review Board of Medstar Research Institute.

2.2. Experimental apparatus

The primary equipment utilized in this study was the Lokomat robotic orthosis manufactured by Hocoma AG (Volketswil, Switzerland). This system is comprised of a treadmill and body-weight support system (Woodway Inc., Waukesha, WI, USA), and two light-weight robotic actuators that attach to the subject's legs (Fig. 1). The Lokomat is fully programmable, including control of knee and hip kinematic trajectories, the amount of assistance the system provides to the patient, and the speed at which the patient ambulates. This high-level dynamic control is achieved by small DC motors and linear ball screw assemblies at the hip and knee joints, tightly synchronized with the timing of the treadmill. The knee and hip joints have position sensors and force sensors that are monitored by the control computer throughout the training. The entire Lokomat assembly resides on a parallelogram structure which in turn is counter-balanced by a large spring. The pretension in the spring is adjusted so that the weight of the Lokomat is compensated for, preventing upward or downward external forces to the subject during training.

For the portion of the study where subjects walked on the treadmill without the Lokomat, knee flexion–extension and varus–valgus angles, as well as hip flexion–extension and abduction–adduction angles were measured using goniometers (XM180, Biometrics Ltd., Ladysmith, VA, USA) while heel contact information was acquired using foot-switches (MA-150, Motion Labs, Baton Rouge, LA, USA). The foot-switch was taped directly to the subject’s heel, while the

Fig. 1. Lokomat® robotic-orthosis (Hocoma AG, Volketswil, Switzerland).
goniometers were taped to the skin using double sided medical tape.

During both Lokomat and treadmill walking, surface EMGs were recorded differentially from the gastrocnemius, tibialis anterior, hamstrings, rectus femoris, adductor longus, vastus lateralis, and gluteus medius and maximus muscles using a Bagnoli-8 EMG system (Delsys, Inc., Boston, MA, USA). Only one leg was instrumented with EMG electrodes since none of the subjects had any gait disorders and walked symmetrically. Knee and hip angles, foot-switch data, and EMG signals were all anti-alias filtered at 500 Hz prior to sampling at 1000 Hz using a 16-bit data acquisition board (Measurement Computing, PCI-DAS 6402, Middleboro, MA, USA) and custom software (Matlab, Mathworks Inc., Natick, MA, USA).

2.3. Protocol

After providing informed consent, EMG electrodes and goniometers were attached to the subject’s leg, which was then wrapped with ace-bandages to ensure the wires did not impede the subject’s gait pattern. After the subject was fully instrumented, they were asked to walk at a self-selected speed on the treadmill for approximately 5 min in order to acclimate to the treadmill. After this acclimation phase, each subject walked at four different walking speeds (0.42, 0.53, 0.64, 0.75 m/s), with the order randomly selected in order to eliminate any bias associated with the order in which speeds were tested. With each change in treadmill speed, the subject walked for 1 min acclimation phase, after which EMG and kinematic data was collected for 60 s.

After walking at the four different speeds on the treadmill, the subject was placed inside the Lokomat described above. The Lokomat linkages were adjusted to the leg lengths of each subject, so that the hip and knee joints of the Lokomat were aligned with those of the subject. The ankle of the subject was held in a neutral position (∼90°) using cloth straps with elastic properties. This was done to replicate the training setup commonly used with individuals with neurological injuries, where the straps are used to assist with dorsiflexion for adequate toe-clearance during swing.

After the device was adjusted to the subject’s geometry, the Lokomat initiated stepping, after which the subject was instructed to try and match the kinematic trajectories dictated by the device. In this sense, the Lokomat was run in a position control mode, where the legs of the Lokomat move through a pre-determined gait pattern. At the onset of walking within the Lokomat, the hip and knee trajectories of the Lokomat were adjusted to best match the joint angles the subject utilized during the treadmill ambulation portion of the study at an intermediate speed. The peak error between the knee and hip trajectories was always within ±10° across the gait cycle. It should be noted that since the kinematics of the subject used during treadmill walking were necessary for adjusting the Lokomat’s kinematic pattern, all subjects first walked on the treadmill followed by the Lokomat.

The subject was first allowed to walk in the Lokomat for 5 min in order to acclimate to the device. After this acclimation period, the Lokomat walking speed was then randomly set to one of the four speeds used during treadmill ambulation, and after 1 min acclimation phase, EMG and kinematic data was collected for a 60-s step sequence. This same procedure was repeated for all four speeds (0.42, 0.53, 0.64, 0.75 m/s). In addition to the experimental signals outlined in Section 2.2, the interaction forces exerted by the subject on the Lokomat were also measured using six degrees of freedom load cells (JR3 Inc, Woodland CA, USA) mounted between the Lokomat and the leg cuffs attached to the subject. Similar to all other signals, the load cell signals were anti-alias filtered at 500 Hz prior to sampling at 1000 Hz using a 16-bit data acquisition board (Measurement Computing) and custom software (Matlab, Mathworks Inc.).

It should be noted that the subject was not provided any body-weight support while walking in the Lokomat as is commonly done when training individuals with neurological injuries. We chose not to use body-weight support since none of the test subjects had any gait disorders and consequently did not require external support. Furthermore, since body-weight support has also been shown to alter muscle activation patterns (Ferris et al., 2001; Finch et al., 1976; Harkema et al., 1997; Dietz et al., 2002; Ivanenko et al., 2002), we did not want to introduce any other biases that could potentially influence the behavior exhibited while walking in the Lokomat.

2.4. Data analysis

Individual stride cycles were determined using the foot-switch data, where each stride was considered the period between successive heel-strikes in the same leg. The muscle activation (EMG) pattern for each stride was then time normalized, expressed as a percentage of the total gait cycle (e.g., 0–100%) and up-sampled using a cubic spline for averaging purposes (Giakas and Baltzopoulos, 1997). EMGs for each stride were then smoothed using a 50-point root-mean-square (RMS) algorithm (Kenney and Keeping, 1962), after which the mean EMG pattern generated for the gait cycle was computed by averaging all the individual stride cycles taken by the subject during the 60-s data collection sequence (n > 20). Finally, each mean EMG trace was normalized to the maximum observed EMG amplitude for each specific muscle across all trials. This allowed for general comparisons across subjects.
After the average EMG profile for each muscle was calculated for all eight trials (4 treadmill, 4 Lokomat), the data was broken up into seven phases as outlined in Table 1 (Perry, 1992). Within each of these phases, the integrated EMG activity was calculated for each muscle.

2.5. Statistical analysis

The amount of EMG activity generated within each of the seven phases of the gait cycle for each muscle was compared between treadmill and Lokomat walking using a two-factor analysis of variance (ANOVA). The fixed factors were the type of walking (i.e. treadmill or Lokomat) and the speed of ambulation while the dependent variable was the integrated normalized EMG activity for each respective phase.

### Table 1

<table>
<thead>
<tr>
<th>Phase</th>
<th>Percent of gait cycle</th>
</tr>
</thead>
<tbody>
<tr>
<td>Initial loading</td>
<td>0–12</td>
</tr>
<tr>
<td>Mid-stance</td>
<td>12–30</td>
</tr>
<tr>
<td>Terminal-stance</td>
<td>30–50</td>
</tr>
<tr>
<td>Pre-swing</td>
<td>50–62</td>
</tr>
<tr>
<td>Initial-swing</td>
<td>62–75</td>
</tr>
<tr>
<td>Mid-swing</td>
<td>75–87</td>
</tr>
<tr>
<td>Terminal-swing</td>
<td>87–100</td>
</tr>
</tbody>
</table>

### Results

#### 3.1. General observations

Despite subjects reporting that walking within the Lokomat was extremely comfortable and felt natural, significant changes in the muscle activation patterns were observed in numerous muscles. Average muscle activation profiles for each muscle group across speeds and conditions are shown in Figs. 2 and 3. Interestingly, there was typically higher muscle activation in the quadriceps muscles (e.g., rectus femoris and vastus lateralis) and the gluteus muscle groups during Lokomat walking than during treadmill walking, while there was often less activation in the gastrocnemius, adductor longus, and tibialis anterior during Lokomat walking. For the gastrocnemius and tibialis anterior muscle groups, the drop in muscle activity may be attributable to the foot lifters placed on the subject’s forefoot which assists ankle dorsiflexion for toe clearance during swing (see Section 2.3 for description).

Mean integrated muscle activity in each of the seven gait phases is shown in Fig. 4, where it is demonstrated that in some muscles, the two walking conditions produce quite similar activation levels while in others, there is consistent over-activity or under-activity, depending on the muscle and phase of the gait cycle. A summary

![Fig. 2. Average normalized gastrocnemius (a), tibialis anterior (b), rectus femoris (c) and vastus lateralis (d) activity across speeds for both treadmill and Lokomat walking.](image-url)
of the statistical differences between the two conditions for each muscle is shown in Table 2.

3.2. Influence of walking speed on EMG activity

The influence of walking speed on the magnitude of EMG activity across the gait cycle was also examined since previous studies have demonstrated a tight correlation between these two factors (Hof et al., 2002; Murray et al., 1894; Nilsson et al., 1985; Ricamato and Hidler, 2004; Shiavi and Griffin, 1983; Shiavi et al., 1987; Yang and Winter, 1985). For the narrow range of speeds tested (0.42–0.75 m/s), there were no statistical differences in the magnitude of EMG for any of the muscles observed in each of the seven phases of the gait cycle, nor where there inter-dependencies between speed and type of walking (e.g., treadmill or Lokomat).

3.3. Interaction forces between subject and robot

Because the EMG patterns of some muscles were different between treadmill and Lokomat walking, we examined the interaction forces between the subject’s legs and the Lokomat to see how closely the subject was able to match the gait patterns prescribed by the Lokomat. As stated previously, prior to collecting any experimental data, we adjusted the kinematic pattern of the Lokomat to best match the kinematic pattern measured at an intermediate speed while walking on the treadmill for that subject so that the step patterns were as comfortable and natural as possible.

We observed that throughout all trials, subjects were able to walk with a similar gait pattern as the Lokomat, however there were always some interaction forces present. In fact, the interaction forces coincided with the measured muscle activity during the gait cycle. Fig. 5 illustrates the mean forces exerted on the upper, middle, and lower Lokomat leg cuffs across all walking speeds and subjects. The left panel depicts the forces along the flexion–extension axis while the right plot is along the abduction–adduction axis. See figure legend for explanation of force effects on leg.

Examining the flexion–extension panel of interaction forces, the thigh cuff illustrates that after heel-strike, the subject drives their leg back into the Lokomat such that the Lokomat resists the desired movement. This behavior correlates with increased muscle activity observed in the hamstrings (Fig. 4). As the leg moves through stance, the direction of the forces reverse and the Lokomat starts to assist the motion of the limb. Surprisingly, there was still significantly more activity in the hamstrings during these phases than during treadmill walking, and also
there was increased activation in the quadriceps. At the end of stance and throughout most of swing, the subject pulls on the Lokomat while at the end of swing, the Lokomat again assists the motion of the subject’s leg. This is
reflected in the large amount of EMG activity in the quadriceps (see rectus and vastus in Fig. 4).

For the lower cuff interaction forces shown in the middle and lower traces in the left column, the subject’s routinely push back on the Lokomat throughout stance and also pull on the Lokomat during swing. In essence the shank is being resisted throughout the gait cycle. It should be noted that while there are interaction forces present despite the subjects trying to match the Lokomat gait pattern, the amplitude of the forces are quite small. This demonstrates their ability to adapt their gait pattern and walk with similar kinematic trajectories as the Lokomat.

For the abduction–adduction forces as shown on the right panel in Fig. 5, throughout most of the gait cycle, the subject exerts a net adduction force on the leg attachment cuffs which corresponds to the significant activity in the adductor longus (Fig. 4). The only portion of the gait cycle in which the subject generates abduction forces occurred during early swing and only for the thigh cuff. Interpretation of these interaction forces are discussed below.

4. Discussion

Comparisons of muscle activation patterns measured during treadmill ambulation with those exhibited while ambulating within a commercially available robotic orthosis demonstrate significant differences in both spatial and temporal properties for most muscles in the lower extremities. The robotic orthosis used in this study, called the Lokomat, restricts leg movements to the sagittal plane and does not allow for substantial hip rotation or listing, all of which are present in normal walking. Yet despite these restrictions, subjects who participated in this study routinely stated that they did not feel uncomfortable or too restricted when walking within the Lokomat but instead felt as though they were walking normally. We hypothesize that after walking
inside the device for just a short period, subjects are able to adapt their gait pattern and learn a new set of motor commands necessary for the restricted movement imposed by the robot (e.g., lack of leg movements along the abduction-adduction axis or pelvic movements such as rotation or listing). This is supported by the small interaction forces between the leg and Lokomat (see Fig. 5).

It is not surprising that differences in the activation patterns of some muscles were found since the device imposes numerous restrictions on the gait pattern. Reducing the degrees of freedom through which the person is allowed to move is necessary for safety reasons, as well as to simplify the complexity of the device so physical therapists can quickly setup the patient. Furthermore, until a wide scale clinical trial of the device of varying complexity is undertaken, it is unclear whether additional degrees of freedom are necessary to promote gains in functional walking ability in individuals with neurological injuries.

4.1. Interpretation of findings

We believe that the observed changes in muscle activity can be explained primarily due to the restriction of leg movements to the sagittal plane and restrictions on pelvic movement. That is, during the swing phase of the gait cycle, subjects normally rotate and list their hips and also abduct their leg to allow the toe to clear the floor. Because the Lokomat limits movement of the pelvis and prevents abduction movement, the subject will exert excessive muscle activity in the quads to help elevate the foot and prevent toe stubbing. In the extreme case, this can be equated to the gait patterns exhibited by a member of a marching band.

In the current study, we did not examine whether subjects could “learn” the Lokomat’s gait pattern over an extended period of time and thereby reduce the amount of coactivation of antagonistic muscles. This may be an important point since training with a device like the Lokomat often involves repeated exposure thereby allowing the subject to adapt and feel more comfortable which may lead to reductions in cocontraction of antagonistic muscles.

Finally, we believe that because the test subjects tended to walk with their feet spread slightly wider than normal to accommodate for the leg cuffs on the Lokomat, this may help explain why there is excessive activation in the adductor longus across the gait cycle. This is a slight artifact with the current study as we tested young healthy individuals with significant muscle bulk. As a result, the leg cuffs that couple the subject to the Lokomat needed to be larger and spread wider than would normally be used in individuals with neurological injuries (e.g., spinal cord injury).

4.2. Failure to induce changes in EMG patterns with speed

While the finding that variations in EMG patterns were not dependent on speed contradicts previous studies (Hof et al., 2002; Shiavi and Griffin, 1983; Shiavi et al., 1987; Yang and Winter, 1985), we believe this is attributed to the extremely slow speeds through which our test subjects walked. That is, in our study, due to limitations with the Lokomat, the speeds subjects walked at ranged from 0.42 to 0.75 m/s. In previous studies that found EMG scaling with speeds, speed ranges were always higher, for example ranging from 0.75 to 1.75 m/s in the study by Hof et al. (2002). At these low walking speeds, it was visually observable that there was significantly more pelvic motion than at normal walking speeds, which could result in decreases in leg muscle activity due to forward propulsion of the legs from pelvis rotation. We also believe that the slow walking speeds resulted in a higher variability in EMG patterns than previously reported (Hof et al., 2002). Reductions in walking speed have been shown to result in both increased pelvic motion and walking variability (Dingwell et al., 2001; Savelberg et al., 1998). As a result of these factors, EMG patterns tended to be smaller and more variable than is commonly observed in normal ranges of walking speeds. It should be stressed that there were observable differences in EMG patterns with walking speed (see Figs. 2 and 3), however the differences were not statistically significant.

4.3. Clinical significance

While the present findings imply that muscle activation patterns exhibited during robotic-assisted gait training are significantly different than those observed during treadmill ambulation, the clinical implications of this are not necessarily negative. That is, during early neurorehabilitation, it is often helpful if not mandatory to reduce the degrees of freedom through which a person can move since neurologically impaired patients can easily become overwhelmed with the amount of tasks to perform. Training in a robotic orthosis allows non-ambulatory patients to start practicing patterned movements through consistent, time-unlimited training sessions earlier in their rehabilitation program. Therefore, simply because the Lokomat restricts movements and alters some muscle EMG patterns should not diminish the fact that the device allows patients to execute mass-practiced movements in a highly consistent manner.

It should be stressed that the findings in this study cannot necessarily be extrapolated to individuals with neurological injuries. Presumably these individuals will
exhibit much different patterns of activity than non-disabled individuals due to the loss of descending motor commands. However the EMG patterns measured in this study do provide a basis of comparison against which the muscle activation patterns exhibited by subjects with neurological injuries while walking in the Lokomat can be compared. New quantitative methods of analyzing muscle activation patterns during gait require normative EMG profiles to evaluate the temporal and spatial properties (Ricamato and Hidler, 2004). The findings in this study suggest that comparing EMG’s demonstrated during robotic-assisted gait training with those collected from healthy subjects walking on a treadmill or over-ground will produce biased results.

4.4. Future directions

While the focus of this study was on the Lokomat robotic-gait orthosis, a number of other devices are already commercially available (Hesse and Uhlenbrock, 2000) or being developed (Reinkensmeyer et al., 2002). Each of these devices work in different ways, however the principle objective of each is the same: to deliver mass-practice gait training to subjects who are unable to ambulate without significant external assistance from physical therapists. Future advances in the robotic-gait orthosis industry may include adding degrees of freedom to the pelvis and the leg, as well as adding complex control strategies to guide leg movements such as adaptive or impedance control. Each of these advances may result in more normative muscle activation patterns and provide new challenges to the patient during neurorehabilitation.

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