FEEDBACK DRIVEN LOCOMOTOR ADAPTATION IN A HUMAN SPINAL CORD INJURY POPULATION

Keith Gordon[1], Ming Wu[1, 2], Jennifer Kahn[1] and Brian Schmit[1, 2, 3]

[1] Rehabilitation Institute of Chicago, Chicago, IL, USA
[2] Northwestern University, Chicago, IL, USA
[3] Marquette University, Milwaukee, WI, USA
e-mail: keith-gordon@northwestern.edu

INTRODUCTION

Spinally transected rats are capable of rapidly adapting locomotor patterns in response to external perturbation (Heng and de Leon, 2007). This finding has important clinical implications for gait rehabilitation following spinal cord injury. We have demonstrated that during stepping, humans with spinal cord injury increase hip extension torque when a dorsiflexor torque is applied about the ankle during stance phase (Gordon et al., 2007). We hypothesized that humans with spinal cord injury would adapt their locomotor patterns (demonstrated by aftereffects) following a short training period of stepping with an externally applied ankle-foot load. Specifically, we expected to see a lasting increase in hip extension torque following training. We also performed catch trials to investigate changes in the control strategy underlying locomotor adaptation.

METHODS AND PROCEDURES

We recorded EMG, kinematics and joint torque from the lower limbs of eight subjects with incomplete spinal cord injury, ASIA C and D, actively performing airstepping (stepping movements performed with 100% body weight support). A Lokomat provided kinematic assistance at the knee and hip joints. Bilateral, powered ankle-foot orthoses were used to mechanically load the ankle and foot. When powered, the orthoses created a dorsiflexor torque (~0.5 Nm/kg) during the stance phase of the step cycle. In the disengaged state, the orthoses allowed free sagittal plane rotation about the ankle.

Subjects performed airstepping for 12 minutes (~300 steps). No load was applied to the subjects' ankle and foot during the first and last 100 steps. During the intermediate 100 steps, subjects received bilateral, ankle-foot stance load. Twelve catch trials, distributed across the three testing conditions, were included to investigate the feed forward / feedback components of the subjects' locomotor strategy. Catch trials during the two no load conditions consisted of loading a single limb during the stance phase of one step cycle. Conversely, catch trials during the stance load condition consisted of removing load unilaterally during a single step cycle.

RESULTS

During the initial no load condition subjects increased hip extension work by ~ 59% when given stance load catch trials (0.35 J/kg) compared to the immediately preceding no load step (0.22 J/kg), ($p < 0.05$) (Fig. 1, 2). Hip extension work also increased significantly ($p < 0.05$), ~160%, during the stance load condition (0.57 J/kg) compared to the initial no load condition (Fig. 1). Hip extension work during the first no load catch trial (0.46 J/kg) was not significantly different from preceding stance load step ($p > 0.05$).

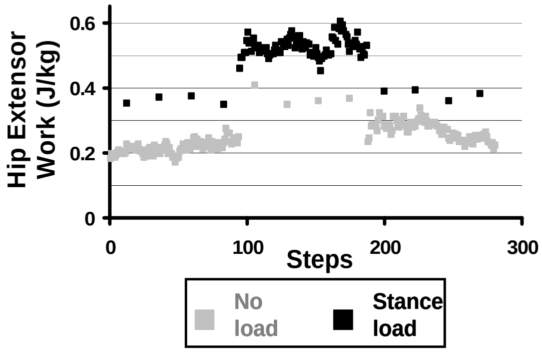

Figure 1. Mean hip extension work increased significantly when load was applied. Changes in the difference between catch trials and baseline may indicate changes in locomotor control strategy.

However, work performed during the stance load step immediately preceding the final no load catch trial was ~58% greater than work performed during the catch trial (0.36 J/kg) ($p < 0.05$) (Fig. 1, 2). Following the transition from the stance load condition to the no load condition subjects decreased hip extension work significantly ($p < 0.05$) (Fig. 1). However, hip extension work initially remained elevated during the final no load condition compared to the initial no load condition (Fig. 1). During the final no load condition, hip extension work performed during the first stance load catch trial (0.38 J/kg) was not significantly different than work performed during the preceding no load step (0.30 J/kg) ($p > 0.05$). In contrast, hip work during the final stance load catch trial (0.38 J/kg) was significantly greater than the preceding no load step (0.24 J/kg) ($p < 0.05$).

DISCUSSION

Results from catch trials performed during the initial no load condition suggest that in spinal cord injury subjects potentially 60% of hip extension work performed during stepping is regulated by feedback from ipsilateral ankle-foot load. In addition, we observed further increases in hip extension work when subjects received multi-step bilateral ankle-foot load. This additional increase can be attributed to several factors including load afferents from the contralateral limb and a progressive increase in reflex response to the repetitive stimulation. These factors may also explain why hip extension work did not return to the no load baseline levels during the catch trials of the stance load condition. Following the stance load condition, subjects' hip extension torque remained elevated for an extended period after ankle-foot load was removed. The change in hip extension work observed during the catch trials vs. no load stepping during the final condition suggests that the elevation in hip extension torque may have been a result of subjects initially adjusting their neural strategy to rely more on feed forward / non-ankle-load related feedback control. These findings may be valuable for improving gait rehabilitation methods.

REFERENCES

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Heng C and de Leon RD (2007). J. Neurosci. 27, 8558-8562.

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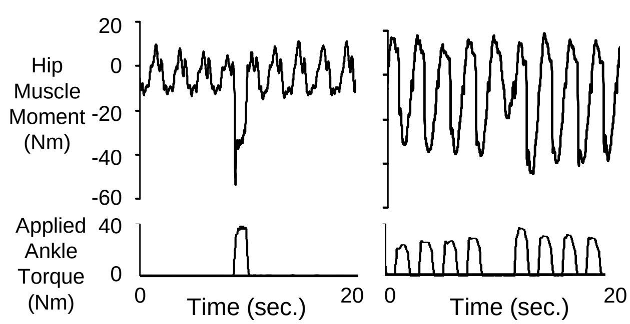

Figure 2. Individual data from the initial no load and stance load stepping conditions.