HARNESS-SUPPORT COMPLIANCE IN TREADMILL TRAINING IN POST-STROKE HEMIPARESIS

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INTRODUCTION

Harness-supported treadmill training, where individuals are partially supported by a harness while stepping is practiced on a motorized treadmill surface, has generated tremendous clinical excitement for its potential to enhance locomotion in individuals with spinal cord injury (SCI) and post-stroke hemiparesis (Dobkin, 1999; Hesse et al., 1995). The technique is believed to be effective because it provides task-specific training and makes the sensory inputs associated with walking more normal for individuals who cannot otherwise walk with proper gait kinematics. Body weight support (BWS) has been accomplished using a number of different methods (i.e., with a winched rope, springs, pneumatic lift, or counterbalancing weights) that permit different degrees of body motion. Compliant support systems permit modest trunk motion while maintaining a relatively stable support force. Stiffer systems exert a more variable force while constraining trunk motion more tightly. Differences in the BWS system used during treadmill walking has been shown to affect vertical ground reaction forces (GRFs) and center of mass movement in neurologically healthy and SCI subjects (Gordon, et al., 2000). In this study, we compared the effect of incremental adjustments in harness-support compliance on 3-D harness forces (vertical, anterior-posterior, and lateral) and vertical GRFs generated by four neurologically healthy subjects and one individual with post-stroke hemiparesis during treadmill walking.

METHODS

Vertical, ant-post, and lateral harness forces applied to the subject were quantified using a custom-made harness support frame instrumented with a 6-axis force/torque sensor (Chen et al., 2001). Vertical GRFs were estimated using insole pressure sensors. Harness-support compliance was set between 0 – 0.855 mm/N by connecting 0 – 3 springs (K = 35.1 N/cm) in series with the support cable. Body weight support was set between 0 - 55% of the subject’s bodyweight. Treadmill speed was set between 0.45 – 1.79 m/s for neurologically healthy subjects and 100 - 140% of self-selected overground walking velocity (0.58 – 0.81 m/s) for the hemiparetic subject. Data was collected from 30-sec walking trials and averaged and normalized to a gait cycle beginning and ending with right heel strike.

RESULTS AND DISCUSSION

In neurologically healthy subjects, vertical harness support was highest during double limb support (when trunk height is lowest) and lowest during midstance (when trunk height is highest) (Fig.1). When harness-support compliance was increased from 0 to 3 springs, mean peak-to-peak fluctuation in vertical harness support decreased from 173.6 (±19.6) N to 36.0 (±10.7) N. Concurrently, vertical GRFs changed from a flat to a normal, double-peaked profile.

In the left-hemiparetic subject, vertical harness support was higher during stance of the paretic left limb, indicating greater reliance on the harness (Fig. 2; from approximately 60 – 20% of gait cycle). When harness-support compliance was increased from 1 to 3 springs, the difference in vertical harness support between the two stance phases decreased, while cycle to cycle variability in ant-post harness forces and vertical GRFs of the paretic limb increased (Fig. 2). The small difference in treadmill velocity between the two trials (0.81 m/s vs. 0.70 m/s) is unlikely to cause the observed differences.

In both neurologically healthy and hemiparetic subjects, changes in harness support tension at low compliance levels accentuated fluctuations in ant-post harness forces but were of more consequence in the hemiparetic subject who generated much higher levels of ant-post harness force (compare Fig. 2 with 1).

SUMMARY

Neurologically healthy and hemiparetic subjects achieved different interactions with the harness. In neurologically healthy subjects, harness-support compliance affected fluctuation in vertical harness support, which altered vertical GRF profiles. In the hemiparetic subject, harness-support compliance affected vertical harness support on the paretic limb, gait cycle variability, and ant-post harness force fluctuation magnitude. Differences in gait mechanics due to harness-support compliance could influence afferent information important to treadmill training. Understanding how abnormal gait mechanics and harness forces interact may lead to improved training parameters for rehabilitation patients.

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ACKNOWLEDGEMENTS

This work was supported by a Whitaker Foundation Predoctoral Fellowship, NIH grant NS17662, and the Rehabilitation R&D Service of the U. S. Department of Veterans Affairs (VA). We also thank Dr. Daniel Ferris for his assistance.

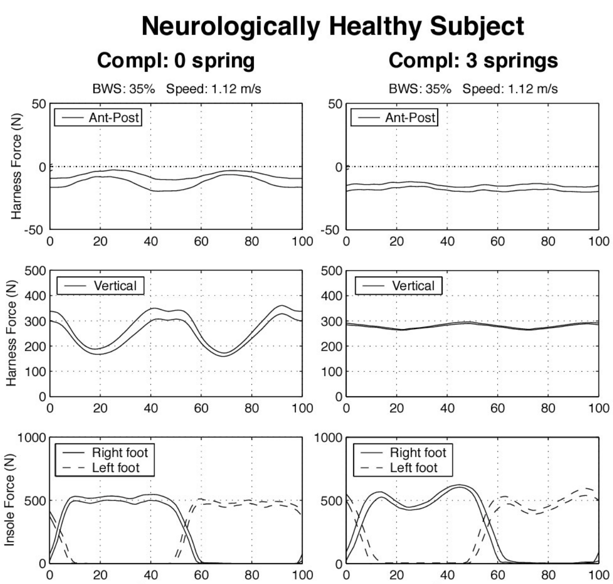

Fig. 1 Normalized trials of a neurologically healthy subject at a support compliance provided by 0 and 3 springs. Region between lines represent ±1SD of mean. Harness forces applied to the subject directed upward and forward are shown as positive.

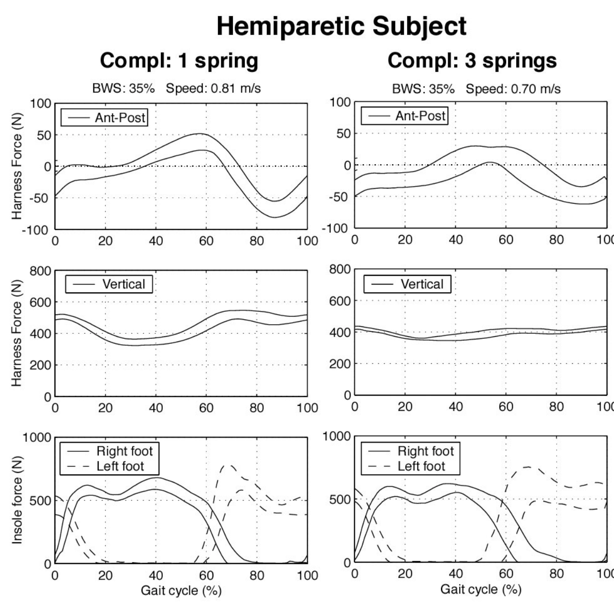

Fig. 2 Normalized trials of a left-hemiparetic subject at a support compliance provided by 1 and 3 springs.