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Loading and knee flexion after stroke: Less does not equal more

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ABSTRACT

It is believed that force feedback can modulate lower extremity extensor activity during gait. The purpose of this research was to determine the role of limb loading on knee extensor excitability during the late stance/early swing phase of gait in persons post-stroke. Ten subjects with chronic hemiparesis post-stroke participated in (1) seated isolated quadriceps reflex testing with ankle loads of 0–0.4N m/kg and (2) gait analysis on a treadmill with 0%, 20% or 40% body weight support. Muscle reflex responses were recorded from vastus lateralis (VL), rectus femoris (RF), and vastus medialis (VM) during seated testing. Knee kinematics and quadriceps activity during late stance/early swing phase of gait were compared across loading conditions. Although isolated loading of the ankle plantarflexors at 0.2 N m/kg reduced VM prolonged response ($p = 0.04$), loading did not alter any other measure of quadriceps excitability (all $p > 0.08$). During gait, the use of BWS did not influence knee kinematics ($p = 0.18$) or muscle activity (all $p > 0.17$) during late stance/early swing phase. This information suggests that load sensed at the ankle has minimal effect on the ipsilateral quadriceps of individuals post-stroke during late stance. It appears that adjusting limb loading during rehabilitation may not be an effective tool to address stiff-knee gait following stroke.

1. Introduction

Following stroke, gait is often marked by decreased knee flexion of the paretic limb during swing phase; a pattern called “stiff-knee gait” (Kerrigan et al., 1991). While the exact cause of stiff-knee gait is unknown, abnormal muscle timing of the quadriceps has been implicated (Kerrigan et al., 1991; Lewek et al., 2007; Waters et al., 1979). Normally, quadriceps activation occurs during the early to mid-stance period of gait; however, individuals post-stroke often demonstrate prolonged quadriceps activity (Kautz and Brown, 1998). The presence of prolonged quadriceps activity into late stance has been associated with stiff knee gait in children with cerebral palsy (Goldberg et al., 2004).

Muscle activity during gait is modifiable by sensory receptors throughout the lower extremity. In particular, force (i.e., load) feedback to the central nervous system (CNS) from Golgi tendon organs (group Ib) serves a critical role in regulating extensor activity duration during stance (Dietz et al., 2002; Hiebert and Pearson, 1999). Prolonged mechanical loading to the ankle extensors, for instance, elicits prolonged muscle activity in the homonymous muscle of decerebrate walking cats (Crone et al., 1988; Hiebert and Pearson, 1999). Humans also appear to exhibit similar positive autogenic force feedback (Dietz et al., 2002; Gordon et al., 2009; Yang et al., 2004), suggesting that the CNS is particularly sensitive to plantarflexor force feedback, and neuromotor output can be directly influenced by these signals (Ada et al., 2010).

Although force feedback appears to excite homonymous muscles during gait (Conway et al., 1987; Duysens and Pearson, 1980), there is evidence for heterogenic inhibitory-force feedback from the plantarflexors to the quadriceps (Ross and Nichols, 2009; Wilmink and Nichols, 2003). This would suggest that increased limb loading (particularly into late stance) would increase the Ib sensory signals from the plantarflexors, to inhibit quadriceps activity during late stance to allow for improved knee flexion. During clinical practice, however, walking on a treadmill with body weight support (BWS) has been used extensively following stroke to facilitate walking recovery (Ada et al., 2010; Barbeau and Visintin, 2003; Visintin et al., 1998). While not shown to be superior to other forms of therapy (Duncan et al., 2011), gait training with BWS has demonstrated improvements in gait speed, endurance,

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and balance (Barbeau and Visintin, 2003; Visintin et al., 1998). Evidence from unimpaired adults (Colby et al., 1999; Dietz et al., 2002; Finch et al., 1991) and individuals post-stroke (Hesse et al., 1999) suggests that the use of BWS will decrease quadriceps activity during early stance. The effect on late stance, however, when plantarflexor load is greatest, remains unknown. In addition, the use of BWS appears to decrease swing phase knee flexion of unimpaired individuals (Finch et al., 1991; van Hedel et al., 2006), representing a potentially unwanted side effect of BWS for individuals post-stroke at a time when knee flexion may already be reduced.

The purpose of this study was to characterize the heterogenic force-feedback response from the plantarflexors to the quadriceps in individuals with chronic (>6 months) stroke. We hypothesized that a simulated ankle load during seated isolated joint reflex testing would decrease knee extensor excitability. We then wanted to determine the presence of a relationship between ankle-load mediated reflex control of the quadriceps and the generation of abnormal quadriceps activity during walking in individuals post-stroke. Based on animal models that manipulated limb load during gait (Hiebert and Pearson, 1999), we hypothesized that walking with BWS would reduce any abnormal late stance/early swing phase quadriceps activity and consequently improve swing phase knee flexion for individuals with chronic (>6 months) stroke. Although all joints in the lower extremity are loaded and unloaded during gait, the ankle extensors are believed to contain the critical Ib afferents to modify gait (Dietz and Duysens, 2000), and thus we expected to observe a direct relationship between plantarflexor-applied, limb load mediated reflex excitability of the quadriceps and the late stance/early swing phase quadriceps activity during gait in individuals post-stroke.

2. Methods

Ten subjects with chronic hemiparesis (>6 months) resulting from a hemorrhagic or ischemic stroke were recruited for testing (see Table 1). Subjects exhibited a variety of walking patterns (Mulroy et al., 2003) but were able to step independently on a treadmill. Subjects were excluded from the study if they responded affirmatively to questions about confounding medical issues (e.g., cardiac arrhythmia, uncontrolled seizures or hypertension) that would preclude the ability to ambulate safely on a treadmill, bilateral stroke involvement, vestibular or cerebellar involvement, or other neurologic and musculoskeletal disorders affecting the legs. Subjective clinical judgment was used to determine the presence of other cognitive or communication problems that inhibited understanding of study involvement. Prior to participation, all subjects signed an informed consent form approved by the University of North Carolina at Chapel Hill Institutional Review Board. All subjects performed the two parts (isolated joint reflex testing, followed by an instrumented gait analysis) of the study during a single session.

2.1. Reflex testing and analysis

Isolated reflex testing was performed to determine the unique contribution of limb load (as it is sensed by the ankle plantarflexors) on knee extensor reflex excitability in a controlled setting (Wu and Schmit, 2006). Prior to testing, active surface electrodes (Motion Lab Systems, Baton Rouge, LA) were applied to the rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), and medial hamstrings (MH) to monitor reflex response. Subjects sat on an isokinetic dynamometer (Humac, CSMi, Stoughton, MA) with the back supported and the hips flexed to 80°. The paretic knee's flexion/extension axis was aligned with the Cybex motor axis and the paretic foot was secured to the footplate of a custom designed ankle loader (see Fig. 1). The ankle loader simulated weight bearing during stance through the production of a dorsiflexion load (Wu and Schmit, 2006). An air powered linear actuator generated and transmitted the load to the footplate. A load cell (MLP-100, Transducer Techniques, Temecula, CA), in series with the actuator, monitored the dorsiflexion load applied to the subject. Dorsiflexion loads were normalized to the subject's body mass, and were set at 0, 0.2, and 0.3–0.4 N m/kg. Some subjects had difficulty tolerating 0.4 N m/kg, so a lower load was used (i.e., 0.3 N m/kg) such that all subjects denied any discomfort during testing. These loads represent a portion of the ankle moments experienced during walking (Lewek, 2011), but are comparable to the passive dorsiflexion moment required to achieve maximum dorsiflexion (Schindler-Ivens et al., 2008). During each loading condition, ramp-hold stretches at the knee joint were performed by passively flexing the knee from 30° to 90° of flexion at 180 and 300°/s. These knee angles are slightly greater than what is observed during 'normal' gait, however, we chose these angles to ensure an adequate stretch of the quadriceps, and to avoid the possibility of hamstring tightness in a more extended position. The speeds were chosen to simulate joint speeds similar to those experienced during gait (Granata et al., 2000), as higher velocity stretches produce higher tonic stretch reflexes (Crone et al., 1988). Therefore, there were a total of 6 randomized conditions (three loads and two speeds). Prior to each trial, subjects pre-activated the hip flexors (approx 10%MVIC) (Lewek et al., 2007) to facilitate the production of RF responses and simulate the hip flexion force that occurs during the stance – swing transition of gait. Subjects were asked to maintain the pre-activated level of effort through the duration of the trial.

During each trial, the knee's angular position, velocity, and torque were recorded from the Cybex. EMG signals were bandpass filtered between 20 and 500 Hz and sampled at 1000 Hz to a personal computer. Analysis of EMG signals was performed with custom software (Labview 2009, National Instruments, Austin, TX). A linear envelope was created by full-wave rectifying the data, followed by low-pass filtering with a 20 Hz, phase-corrected, 8th order Butterworth filter. To avoid values greater than 1.0, the linear envelope was normalized to either the peak recorded during an MVIC collected before performing the reflex testing, or the peak muscle activity recorded during the gait analysis (Rudolph et al., 2001).

Table 1
Subject demographics.

Subject	Sex	Age	Paretic side	Time since stroke (months)	Treadmill walking speed (m/s)	AFO	Assistive device
1	M	45	L	55	0.6	n	y
2	F	61	L	21	0.5	n	y
3	M	72	R	22	0.3	y	y
4	F	66	R	31	0.45	n	y
5	M	51	L	89	0.7	y	y
6	M	57	R	73	0.8	n	n
7	F	62	R	14	0.6	n	y
8	M	68	L	195	0.4	n	n
9	F	56	L	356	0.8	n	n
10	M	63	R	8	0.65	n	y