THE INFLUENCE OF INCREASING STEADY-STATE WALKING SPEED ON MUSCLE COORDINATION IN BELOW-KNEE AMPUTEES

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INTRODUCTION

Previous studies have shown that below-knee amputees exhibit greater energy expenditure and bilateral asymmetry relative to non-amputees at their self-selected walking speed (e.g., Hafner et al., 2002). These differences are attributed to the functional loss of the ankle plantar flexors, which are critical to providing body support, forward propulsion, and swing initiation in non-amputee walking (Neptune et al., 2004). Identifying how amputee muscle coordination changes in response to changes in task demands, such as increased walking speed, will provide insight into the compensatory mechanisms used by amputees. Previous studies have examined muscle activity during amputee walking, but have either analyzed a small set of muscles (typically < 3) or not compared their results with control subjects under the same conditions. Furthermore, no study has examined amputee muscle activity over a wide range of walking speeds. The purpose of this study was to examine bilateral muscle activity across a wide range of walking speeds to identify changes in muscle coordination in below-knee amputees.

METHODS AND PROCEDURES

Fourteen below-knee amputees and 10 control subjects walked at 0.6, 0.9, 1.2 and 1.5 m/s. Kinematic, ground reaction force (GRF), and electromyographic (EMG) data were collected using a Vicon motion capture system (Oxford Metrics, Inc.). EMG data were collected using surface electrodes from eight intact leg muscles including the tibialis anterior (TA), medial gastrocnemius (GAS), soleus (SOL), vastus lateralis (VAS), rectus femoris (RF), biceps femoris long head (BF), gluteus medius (GMED), and gluteus maximus (GMAX), and from five residual leg muscles including VAS, RF, BF, GMED, and GMAX. Integrated EMG (iEMG) magnitude was calculated over the gait cycle and within three subphases including braking (~0-50% stance), propulsion (~50-100% stance) and swing. Gait cycle and subphase iEMG magnitudes were normalized by the gait cycle iEMG value at 1.5 m/s. This method of normalization was performed at the individual subject level, and individual values were averaged for each group.

Statistical analyses compared iEMG values and included five, three-factor (group, leg, speed) repeated measures ANOVAs for the upper-leg muscles and three, two-factor (group, speed) repeated measures ANOVAs for the intact and control leg ankle muscles. The same analysis was repeated for the gait cycle and each subphase. When significant differences were found, Bonferroni pairwise comparisons were used to determine which values were significantly different ($p \leq 0.05$).

RESULTS

The most notable differences occurred within the residual leg muscles. During braking, significantly higher activity was found in the residual leg VAS (all speeds compared to the intact leg) and BF (all speeds compared to the intact leg, and at 1.5 m/s compared to the control leg) (Fig. 1).

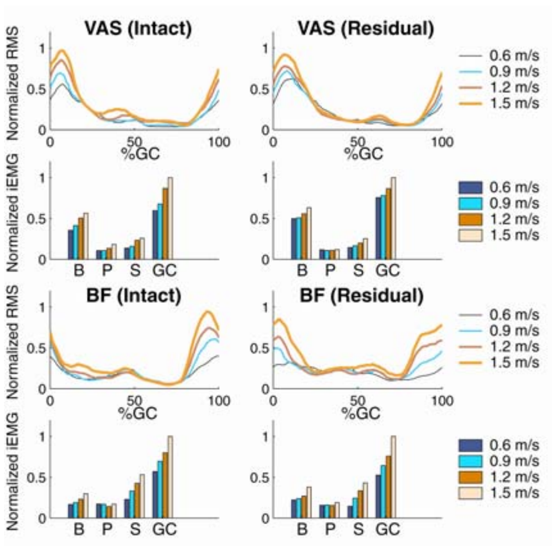

Figure 1. Intact and residual leg EMG profiles and iEMG magnitudes for VASL and BF muscles within Braking (B), Propulsion (P), Swing (S), and the gait cycle (GC).

Other significant findings included increased residual leg RF activity during braking (0.9, 1.2, and 1.5 m/s compared to the intact leg, and at 0.9 and 1.2 m/s compared to the control leg) which resulted in a more systematic response to walking speed compared to the intact and control legs. Similar to the residual leg VAS, the residual leg GMAX exhibited significantly higher activity during braking compared to the intact leg GMAX at 0.6 and 0.9 m/s. Also, GMED activity in all legs was largely insensitive to changes in walking speed.

DISCUSSION

The heightened and prolonged residual leg VAS and BF activity (Fig. 1), especially at the slower speeds, suggest a need for increased stability in the absence of the plantar flexors (e.g., Rietman et al., 2002). In addition, the heightened and prolonged residual leg BF activity may help to reduce residual leg braking. Previous modeling and simulation studies of non-amputee walking have shown BF to contribute positively to the anterior/posterior GRF throughout stance (e.g., Neptune et al., 2004). Thus, increased activity of this muscle in early stance could provide propulsion, and therefore reduce the net braking impulse.

The higher activity of the residual leg RF during braking at 0.9 and 1.2 m/s, compared to control legs, suggests that it may be transferring more energy from the leg to the trunk to provide additional body support (e.g., Neptune et al., 2004). In addition, the residual leg GMAX, along with the residual leg VAS muscles, may provide increased body support in the absence of the ankle plantar flexors (e.g., Neptune et al., 2004). The insensitivity of GMED to walking speed is consistent with its primary role to provide body support (Anderson and Pandy, 2003).

In summary, most amputee EMG patterns across legs were similar to the control subjects and systematically increased with speed. The most notable differences were observed in the residual leg BF, VAS, RF and GMAX muscles, which exhibited increased output, especially at slower speeds, and appear to provide needed body support and propulsion in the absence of the plantar flexors.

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