ASSISTIVE ANKLE POWER PROMPTS MUSCULAR COMPENSATIONS DURING GAIT

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INTRODUCTION

Muscular compensations by both intact and residual limbs have been described for individuals walking with passive elastic ankle prostheses [1, 2]. These devices can only supply about half the power normally provided by the plantarflexors in the intact human ankle. A commercially-available, powered ankle-foot prosthesis that supplies plantarflexor power with each step has shown promise in improving walking economy and normalizing ankle kinematics and kinetics [3, 4]. However, the neuromuscular adaptations that drive these changes are not yet understood. Characterizing these adaptations will help clarify the mechanisms by which powered devices improve walking economy and may inform future device design. Therefore, the purpose of this study was to characterize the muscular compensations exhibited in response to walking with various amounts of prosthetic ankle power.

METHODS

Two (2) males with trans-tibial amputation were fitted with the powered ankle prosthesis (BiOM T2 Ankle System, BiOM, Inc., Bedford, MA) by a manufacturer-certified prosthestist. Subjects were experienced prosthesis-users (44 and 15 yrs post-amputation; ages 59, 24 yrs old; BMI: 28.9, 26.9 kg/m^2, respectively). Testing consisted of treadmill walking at five ankle power settings in random order: prosthetist-chosen, 0%, 25%, 75% and 100% power (0P, 25P, 75P, 100P). Metabolic cost of transport was measured using a portable gas analyzer. Surface electromyography (EMG) was recorded from gluteus medius (GM), vastus lateralis (VL), rectus femoris (RF), medial/lateral hamstring (MH/LH), medial/lateral gastrocnemius (MG/LG) and tibialis anterior (TA) bilaterally where possible. Treadmill belt speed was set based on leg length (1.16, 1.23 m/s). The first power setting condition was preceded by a 10-minute accommodation period and subsequent accommodation periods were five minutes each. After accommodation, a 3-minute sample of expired air and a 30-second sample of EMG were acquired. Cost of transport was energy expenditure normalized to bodyweight and distance traveled. After visual inspection of raw EMG signals, linear envelopes were generated by full-wave rectifying and applying a low-pass filter (6 Hz). The resultant envelopes were normalized to peak EMG from the prosthetist-chosen settings. Muscle EMG magnitude variables included peak amplitude (pkEMG) and the integral of muscle activity per gait cycle (iEMG). Muscle co-contraction was assessed for 2 muscle pairs: VL-LH and MG-TA. Co-contraction indices equaled the agonist iEMG divided by antagonist iEMG. For analysis, the gait cycle was divided into discrete phases as outlined by Seyedali et al. [5] (for VL-LH: early mid-stance and late swing, for MG-TA: early stance, late stance and early swing).

RESULTS AND DISCUSSION

Metabolic cost of transport tended to decrease with increasing levels of ankle power (means in J/Nm; 0P = 0.396; 25P = 0.427; 75P = 0.375; 100P = 3.57). For the intact limb, the maximum prosthetic ankle power (100P) was associated with large, consistent decreases in EMGpk and iEMG in muscles of the intact limb, particularly the hamstrings, quadriceps and gluteus medius muscles (Table 1). Increased prosthetic ankle power was associated with inconsistent changes in muscle co-contraction indices (Figure 1). In general, the muscles of the intact limb were less active when additional ankle power was supplied (Figure 2). The largest decreases in muscle activity were seen for the intact limb, which suggests that much of the energetic gains provided by the powered ankle prosthesis are accomplished by reducing demand on intact limb muscles crossing the hip and knee.

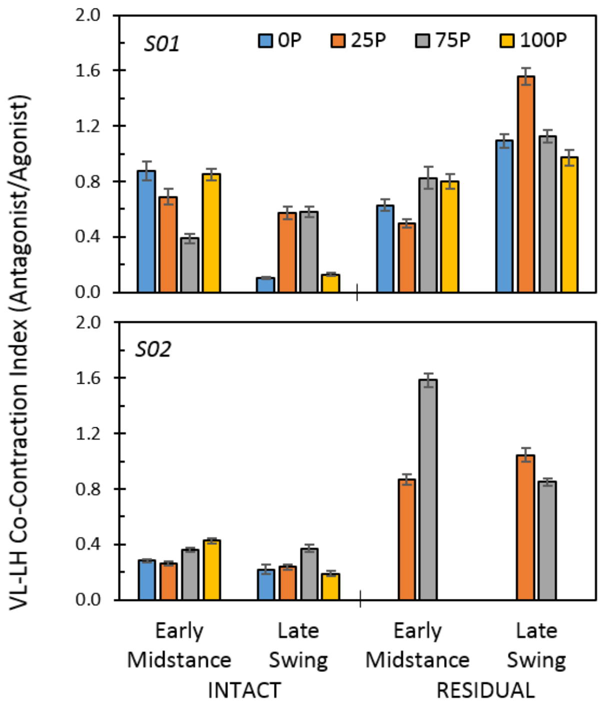

Figure 1. With increased power, co-contraction for residual limb VL-LH increased in early mid-stance and decreased in late swing (means ± SE). Residual EMG unavailable for S02 at 0P and 100P.

For the residual limb, modest-sized decreases in pkEMG and iEMG occurred for hamstrings (Table 1). Changes in other residual limb muscles were small and inconsistent in direction. Persons using passive elastic ankle-foot prostheses walk with greater residual limb hamstring and hip flexor activity [1, 2]. The addition of ankle power in the present study appeared to decrease compensatory hamstring activity. For the residual limb, VL-LH co-contraction was greater than for the intact limb and tended to increase in early stance with increasing ankle power (Figure 1). Co-contraction on the prosthetic side may indicate a limb stiffening strategy adopted to increase stability early in stance.

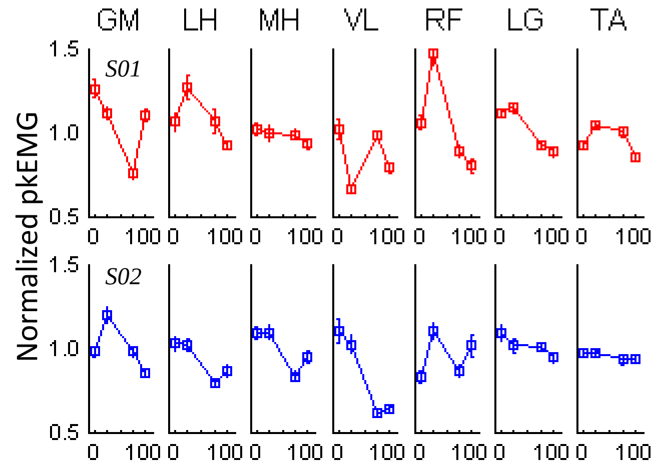

Figure 2. Peak EMG decreased for intact limb muscles of both subjects with increasing power (means ± SE).

CONCLUSIONS

Data from the two individuals tested in the present study suggest that reductions in energy cost of walking with a powered ankle prosthesis may stem from decreased muscular effort of the intact limb. Additionally, residual limb hamstring muscle compensations appear to decrease when using a powered ankle. Further work with more subjects will clarify the mechanisms by which ankle power can reduce energy expenditure during gait.

REFERENCES

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Table 1: Sizes of change (Cohen's d) in peak and integrated EMG for 0P vs. 100P. Large negative effects are shaded and represent decreased muscle activity with the addition of more prosthetic ankle power (100P).

	Subject	GM	RF	VL	LH	MH	LG	TA
Intact iEMG	*S01*	-0.3	-1.3	-0.5	-0.8	-1.4	-1.2	-0.6
	S02	-0.8	0.8	-2.0	-1.8	-1.3	0.2	-0.2
Intact pkEMG	*S01*	-0.8	-1.3	-1.1	-0.7	-0.5	-1.7	-0.7
	S02	-0.9	0.7	-2.1	-0.8	-0.8	-0.7	-0.5