Muscle Co-contraction Strategies Using Dynamic Optimization and EMG While Walking on Uneven Surfaces

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Summary

Humans are unstable in the mediolateral direction, and this instability worsens on uneven surfaces. To compensate, the neuromuscular system utilizes several strategies, including muscular co-contraction. Most studies quantify co-contraction using electromyography (EMG) data only. However, muscle force and EMG are not always linearly correlated. This paper aims to examine muscular co-contraction strategies used on mediolaterally uneven surfaces and compare co-contraction measures derived from muscle moments using an EMG-constrained dynamic optimization procedure (EMG-Track) with those derived from EMG data only (EMG-Based).

Introduction

Muscular co-contraction, defined as the simultaneous activation or contraction of agonist and antagonist muscles, may be necessary for maintaining stability while walking on uneven surfaces [1]. Typically, co-contraction measures involve using EMG data only (EMG-Based). Although EMG is used for convenience, its relationship to muscle force is not always linear [1,2]. An alternative approach for calculating co-contraction is using muscle force estimations obtained through dynamic optimization, particularly muscle moments. Muscle forces are often estimated using efficiency-based objective functions, but EMG-constrained approaches integrate participant specific EMG data [2] and in this study is referred to as an EMG-Track method. The purpose of this study was to compare co-contraction strategies on mediolaterally uneven surfaces using EMG-Based and EMG-Track measures. It was hypothesized that an EMG-Track method will detect differences in co-contraction across footwear conditions and walking speeds with lower magnitudes compared to an EMG-Based method.

Methods

18 healthy participants walked on a force-instrumented split-belt treadmill at 0.8 m/s and 1.6 m/s. Participants walked in three 3D-printed custom-made footwear: 1) 1 cm medial-posterior ridge (MPR), which limited pronation and 2) 1 cm lateral-posterior ridge (LPR) that caused additional pronation, and a no ridge condition (NR) (Figure 1). Muscle excitations and estimated forces were determined for the gastrocnemius medialis (MG), gastrocnemius lateralis (LG), soleus (SOL), tibialis anterior (TA), peroneus longus (PL) and peroneus tertius (PT). Co-contraction of the PL/TA, PL/SOL and PL/MG were compared using the EMG-Based and EMG-Track co-contraction indices.

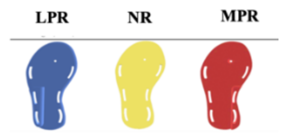

Figure 1: 3D-printed footwear conditions.

Results and Discussion

Overall, the magnitude of co-contraction predicted through EMG-Track was significantly lower than EMG-Based (p<0.05) across all conditions (Figure 2). EMG-Track was more sensitive to detecting differences in co-contraction across different footwear conditions, while EMG-Based was unable to detect the same differences that EMG-Track was able to. EMG-Track was also better at detecting differences in co-contraction across both speeds in the LPR condition.

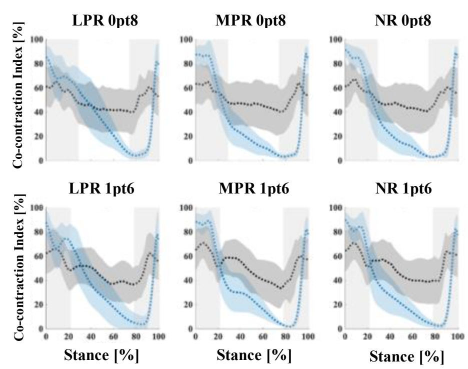

Figure 2: Time-varying co-contraction index of PL/TA during the stance phase. Blue and black lines represent co-contraction for EMG-Track and EMG-Based, respectively. Shaded areas indicate standard deviation. Each stance phase is divided into early (grey), mid (white), and late (grey) stance.

Conclusions

Our results suggest that the EMG-Track method is more valuable when assessing the effects of co-contraction during gait and unstable surfaces. It may also be beneficial for certain populations, such as older adults, where co-contraction is used to assess therapeutic interventions prescribed.

Acknowledgments

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References

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