COMPENSATORY MUSCLE CONTROL STRATEGIES WHEN WALKING WITH A CUSTOMIZED PD-AFO

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INTRODUCTION
Passive-dynamic ankle-foot orthoses (PD-AFOs) use the rotational stiffness of the brace to provide plantarflexor assistance during the stance phase of gait, controlling the tibia as it rotates over the stationary foot. We have developed a novel customization and manufacturing framework for such PD-AFOs and demonstrated the feasibility of rapidly fabricating these fit-customized braces [1]. While the rotational stiffness of these PD-AFOs can be experimentally determined, the optimal stiffness is unknown due to the lack of understanding of how the stiffness influences the compensatory control strategy used when walking in a PD-AFO.

There are a multitude of compensatory control strategies that can be used to walk in a customized PD-AFO. For example, a pure substitution paradigm may be employed where the PD-AFO stiffness directly replaces function of the uniarticular plantarflexors. In other words, the contribution of these plantarflexors to the ankle joint moment decreases by the same amount as is contributed by the PD-AFO stiffness, resulting in no change to the ankle joint moment.

The compensatory strategy employed by the body when walking with a customized PD-AFO of a specific stiffness is currently unknown. Therefore, this study aimed to use a combination of experimental and simulation techniques to identify the compensatory muscle control strategy used when walking in a customized PD-AFO with a known stiffness in order to better understand the mechanisms used to walk with PD-AFOs.

METHODS
Experimental Data: Torso, pelvis and bilateral lower extremity movement analysis data were collected as a healthy subject (age: 24 yrs, height: 1.62m, mass: 63.6 kg) walked overground in two conditions: without and with a customized PD-AFO. The PD-AFO was customized for and worn on the subject’s right leg. The subject wore shoes bilaterally in the normal condition and on the contralateral limb in the AFO condition. Kinematics of the PD-AFO cuff were recorded separately from the shank in the AFO condition. Kinematic and kinetic data were filtered at 6 Hz and 25 Hz, respectively, using a zero-lag low-pass Butterworth filter. The right ankle angles and joint moments during right stance were calculated using an inverse kinematics approach in Visual 3D (C-Motion Inc., Germantown, MD, USA), and these data were compared between the normal and AFO conditions.

Musculoskeletal Models and Simulations: In OpenSim, a lower-extremity musculoskeletal model, with 23 degrees of freedom and 54 muscle actuators, was scaled the subject’s anthropometric measurements [2].

The characteristics of the customized PD-AFO were incorporated into the normal musculoskeletal model for the AFO condition. This PD-AFO-integrated model added two bodies, a footplate and a strut-cuff, to the right lower leg (Fig. 1). The footplate body was welded to the calcaneus. The strut-cuff body was connected to the tibia via a custom joint that only allowed translation of this body along the tibia’s longitudinal axis. The two bodies were connected at their origins, the ankle joint center, by a six degree of freedom (DOF) bushing force, which mimicked a spring with user-prescribed passive stiffness values for each DOF. The stiffness for the dorsi/plantarflexion DOF was set to 3.0 Nm/deg to match the stiffness of the customized PD-AFO used in the experimental AFO condition. All other rotational and translational DOFs of the bushing force were set to zero, as the other DOFs were controlled by joint definitions and the PD-AFO strut-cuff body experimental kinematics.
Using OpenSim v. 2.0.2, a quasi-static forward dynamic simulation tracked experimental gait data under the normal and AFO conditions with the corresponding (normal or PD-AFO-integrated) musculoskeletal model. Predicted muscle activity that drove the simulations was calculated through a computed muscle control optimization scheme [3]. Simulated and experimental right ankle angles were compared to assess tracking of the experimental kinematics. Right medial gastrocnemius (MGAS) and soleus (SOL) activations were examined during right stance for both conditions. Additionally, MGAS, SOL and bushing force moments about the right ankle were examined for the AFO condition.

RESULTS AND DISCUSSION
The subject walked with a decreased ankle range of motion (Fig. 2a) and showed a premature increase in the plantarflexion moment (Fig. 2b) during the AFO condition. Both simulations successfully tracked the experimental ankle kinematics (Fig. 2a).

The simulation predicted that PD-AFO stiffness substitutes for all SOL activity during stance (Fig. 3a). However, the compensatory strategy was not a pure substitution paradigm, as the ankle joint moment differed between the two conditions (Fig. 2b). Furthermore, these changes in net ankle joint moment could not be account for entirely by the PD-AFO stiffness. The bushing force provided an ankle moment proportional to the degree of ankle dorsiflexion, supplying a constant moment during midstance and only accounting for 11% of the peak plantarflexion moment (Fig. 3b). Instead, the ankle moment changes were primarily reflected in altered MGAS activity in the AFO condition (Fig. 3a).

CONCLUSIONS
These results indicated that, for this subject, a PD-AFO with a stiffness of 3.0 Nm/deg induced a complex compensatory muscle control strategy involving changes in both joint kinematics and muscle function. Future work will aim to identify the compensatory strategies that arise when walking in customized PD-AFOs with a range of stiffness values in order to gain insight into the spectrum of compensatory mechanisms. Ultimately, this approach may enable optimization of the PD-AFO design to achieve enhanced function for patients with a range of impairments.

REFERENCES

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