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, FA resistance applied during self-driven walking resulted in increased propulsive-force output of healthy-nonimpaired individuals with accompanying biomechanical changes that facilitated greater limb propulsion. Future rehabilitation interventions for neurological populations may be able to utilize this principle to design task-specific interventions like progressive strength training and workload manipulation during aerobic training for improving walking function.

e , Walking, Fore-aft resistance, Propulsion, Biomechanics, Treadmill-interface nonimpaired

I d

Walking is a complicated motor task that requires generation of lower-limb muscle forces that both propel and speed (10-m walk test, mean speed = 1.2 m/s (SD = 0.03)) prior to study participation.

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this environment are similar to typical treadmill walking [31, 34].

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For this experiment, we selected a target walking speed of 1.0 m/s +/− 0.2 m/s i.e., target speed range, to account for the sinusoidal nature of normal walking. We selected 1.0 m/s as it is close to the comfortable overground speed of healthy-nonimpaired adults, and our previous research involving walking in the KineAssist has demonstrated that a slightly slower than comfortable speed is desirable for quick adjustments to this walking environment [18, 34, 36, 37]. However, in the future we could scale the FA resistance levels to any selected speed based on the system's force-velocity relationship (see Eq. 1 below). Similar to the Gottschall and Kram study [18], we chose four FA resistance levels (10, 15, 20, and 25%) taken as percentages of a participant's vertical body weight. We used an algorithm that accounted for the participant's vertical body weight, interface control parameters, and the system's force-velocity relationship to calculate their target resistance levels. Thus, we normalized FA resistance by body weight for the same target speed using.

$$b = y - mx \tag{1}$$

Where b is the fore-aft resistance (Newtons) to move the belt at an intended speed, y is the percentage of vertical body weight (Newtons) needed to maintain treadmill belt movement at 1.0 m/s, m is the sensitivity constant of the interface (set for all participants at 50 N-sec/m that allowed quick response of the belt with minimum delay (0.01 m/s)), and [illegible]

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We collected individual-limb GRFs via the Bertec instrumented treadmill, with kinetic data sampled at 1000 Hz. We also collected 3D kinematic data, sampled at 100 Hz, using an eight-camera Qualisys motion capture system (Qualisys Inc., Gothenburg, Sweden). We used a custom 36 passive-reflective markers setup with each marker being 1-cm in diameter. Markers were placed bilaterally as follows: acromion processes, manubrium of sternum, thoracic spine level with sternum, ASIS, PSIS, sacrum, midline and lateral thighs, midline and lateral shanks, lateral malleoli, first and fifth digits, and calcaneus (see example marker set-up in Fig. 1) [35]. We collected the real-time velocity of each treadmill belt and forces applied to the pelvic-mechanism force transducers (100 Hz) using our treadmill-interface's custom software.

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(repeated across resistance levels) for all secondary spatio-

(Fig. 5

generation [31, 34] due to the pelvic-mechanism holding participants in place and limiting forward-backward translation of the COM that occurs during typical walking. This effect coupled with requirements of walking against FA resistance may have further reduced braking. Walking against FA resistance is similar to uphill walking [25, 42], which has also shown reductions in braking-force generation, possibly due to increasing demands of raising the COM and earlier need for propulsion generation to maintain target speeds.

Regarding limb angle changes, participants increased their trailing limb angles and decreased their leading limb angles at higher FA resistance levels. Several studies have indicated that an increase in trailing limb angle is a strategy to increase propulsive-force generation [28, 45–48], while reduction in leading limb angle is also indicative of participants trying to quickly get the limb into a more posterior position to propel the COM forward. We believe that such a strategy enabled participants to increase rate and magnitude of propulsive-force generation to meet the demands of greater resistance and maintain walking speed, as stance time and stride time did not significantly change across conditions. These findings are consistent with studies that highlight how increases in trailing limb angle are associated with increases in propulsion needed to attain faster walking speeds [26, 38, 41, 49].

At the individual joint level, walking against greater FA resistance resulted in increased ankle and hip power generation with little to no changes in knee power generation. Collectively, these changes indicated use of an ankle and hip strategy to attain target walking speeds against higher FA resistance levels [24, 42, 50]. Visual analysis of joint moments (not reported here) also revealed increases in positive hip joint moments at higher resistance levels, with minimal to little change in ankle and knee joint moments. This suggests that an increase in ankle joint angular velocity facilitated the increased ankle power generation, while an increase in hip joint force production (moment) facilitated the increased hip power production. This strategy implies that perhaps the larger hip muscles were best suited to lend themselves to the increased demands of fore-aft limb loading to maintain target speed inside the treadmill interface. It has been reported that positive hip joint powers are known to increase significantly over the ankle at faster walking speeds [51, 52]. We found similar joint changes in our study, possibly to move the limb in a position directly underneath and behind the body to increase forward propulsion of the COM. Additionally, we visually noted an absence of hip joint power absorption that typically occurs in terminal stance and is associated with stretching of hip proprioceptors to facilitate offloading to initiate swing. This lack of negative hip work during terminal stance might indicate that walking against FA resistance created a

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We demonstrated that walking against FA resistance, applied by a robotic system that allowed people to control their own speed, proportionately increased fore-aft limb loading without significant changes in vertical limb loading. The experimental environment of the robotic treadmill interface enabled us to manipulate the fore-aft loading demands during stance while participants controlled their walking speeds. Our results suggested that FA resistance can be applied in environments that allow self-controlled walking in order to increase propulsive force output of healthy-nonimpaired individuals.

F

Future studies will evaluate whether this approach may be a useful rehabilitation application, especially for individuals who may have difficulty walking on inclines or at fast speeds. For example, this approach could be used for ambulatory individuals with hemiparesis both as a measurement tool to assess functional strength related to walking (i.e., maximum propulsion capability or propulsion “reserve”), and also serve as a training environment for individuals who are able to increase propulsion under increased FA loading. Individuals poststroke may generate

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