LOW-PROFILE WEARABLE SUIT FOR SENSING HUMAN DYNAMICS

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Introduction: The current state of the art for collecting biomechanical data of humans is to use high precision motion capture systems and force plates. While these systems are very accurate, their drawbacks include inaccessibility, inability to study certain movements in their correct contexts, and significant data processing to achieve results. The field has developed alternatives that utilize wearable technology and developed approaches including the XSens IMU based motion capture system and Harvard's strain sensor-based suit, which can characterize the kinematics of human motion [1][2]. However, no wearable system to date is capable of characterizing dynamics of human motions.

The ability to accurately quantify ground reaction forces as well as kinematics with low-profile wearable technology would enable dynamics data to be collected in many new contexts outside of lab environments. The development of such a sensing suit would be an enabling technology for exoskeleton controllers which are often in a torque control paradigm. Furthermore, real-time feedback could be employed to help users minimize potentially dangerous joint loading consequences of their motions, whether via active exoskeleton assistance, or through sensory feedback to the user, to make them cognizant of dangerous postures. We present and characterize a wearable biomechanics sensing suit that incorporates pressure sensing insoles as a means of characterizing ground reaction forces.

Methods: The Second-Skin is a wearable sensing suit consisting of compression pants and a modified running backpack. Onboard are eight total sensors, six inertial measurement units (IMUs), and two pressure sensing insoles. The IMUs are 3DM®GX5-AHRS series IMUs from Parker Hannifin Corporation©, which are industrial grade precision IMUs that are small and lightweight, and the pressure sensing insoles are OpenGo™ insoles from zFlo. Each IMU is placed on a different segment of the body (Figure 1A), which allows for each segment of the body to be measured independently. Onboard the backpack a NVIDIA® Jetson Nano handles communication and syncing between all the sensors via ROS. Each IMU connects to the Jetson Nano via serial USB and can send its linear accelerations, angular velocities, and Euler angles. Additionally, each Insole connects to the Jetson via Bluetooth, and can send its vertical ground reaction force, center of pressure, linear accelerations, and Euler angles. Lastly, the compression pants and insoles can be easily swapped out for different sizes, and the running backpack can be easily adjusted, allowing the Second-Skin to be correctly fitted to most people.

A simple experiment was performed using the Second-Skin to validate its ability to measure segment angles (sagittal thigh and shank) and vertical ground reaction forces compared to conventional motion capture. One subject performed level ground walking for 45 seconds on a Bertec® instrumented treadmill while being recorded by both the Second-Skin and a Vicon® motion capture system (ground truth). RMSE of kinematic angles were calculated between these systems for the entire duration of the trial. Likewise, the vertical ground reaction forces on the left foot and right foot were measured by the instrumented treadmill and the insoles, and RMSE was calculated between these values for the entire 45 second trial.

Results & Discussion: RMSE of the sagittal plane segment angles of the left shank, right shank, left thigh, and right thigh were .095, .109, .173, and .178 (rad), respectively (Figure 1C). It is possible that the higher error of thigh segment angles relative to shank segment angles is due to the placement of the IMU being further from bone, and thus more affected by changes in soft tissue. RMSE of the vertical ground reaction forces of the left insole and right insole were 31.38 and 36.56 (N), respectively (Figure 1C). IMU measurements were able to be recorded at a rate of 200hz, and insole data was able to be recorded at a rate of 50hz, which is sufficiently fast enough to do sensing for wearables [3].

Significance: The results indicate it may be possible to characterize dynamics of human motion with wearable technology. Doing so would potentially extend the practice of biomechanics research to new environments, and facilitate improvements in the effectiveness of wearable robots.
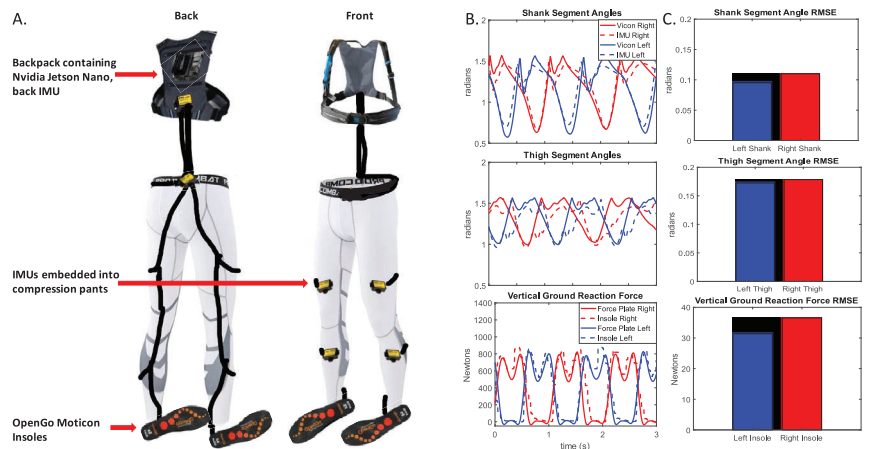

Figure 1. A: Conceptual diagram of the Second Skin. **B:** Timeseries data of recorded shank segment angles, thigh segment angles, and vertical ground reaction forces for three seconds of walking trial. **C:** RMSE between measurements from the Second Skin and measurements from Vicon motion capture system of entire walking trial.

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PLANTARFLEXOR WEAKNESS AND LESSER ACHILLES TENDON STIFFNESS ASSOCIATE WITH GREATER VULNERABILITY TO WALKING BALANCE PERTURBATIONS

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Introduction: Falls among older adults are a significant public health concern and can result in severe injuries with high healthcare costs. Intuitively, the ability to accommodate the instability elicited by walking balance challenges that could precipitate a fall can be impaired through age-related declines in sensory acuity and neuromuscular integrity. Foremost, sensory information from muscle stretch receptors requires rapid transmission of an unanticipated change in joint position. Age-related differences in sensory transmission are well-documented. However, tendon stiffness also decreases with age, which could delay and reduce the velocity of muscle spindle stretch and slow balance perturbation detection. Older adults also experience age-related decreases in muscle strength and power. Thus, even with requisite detection of a balance perturbation, weakness is likely to hinder older adults' ability to produce the requisite forces needed to mitigate instability. Ultimately, the association between tendon stiffness, muscle strength, and vulnerability to walking balance perturbations is not well understood [1]. Treadmill-induced slip perturbations emulate a situation in which falls among older adults are more likely. Moreover, given their proximity to the onset of those perturbations, factors associated with the integrity of muscle-tendon units spanning the ankle – namely Achilles tendon stiffness (k_{AT}) and plantarflexor muscle strength – are first line defenses against instability. Thus, the purpose of our study was to determine whether k_{AT} and plantarflexor strength were associated with vulnerability to treadmill-induced slip perturbations across a cohort of younger and older adults. We first hypothesized that older adults would exhibit lesser kAT and reduced plantarflexor strength compared to younger adults. We also hypothesized reduced k_{AT} and plantarflexor weakness would correlate with a greater vulnerability to perturbations. Data in support of our second hypothesis would provide evidence for potentially-modifiable factors and thus targets for intervention to mitigate falls risk.

Methods: 21 younger (21.7 ± 2.0 years, 66.4 ± 8.6 kg) and 22 older (74.0 ± 6.0 years, 69.5 ± 18.6 kg) adults participated. We first quantified subject's plantarflexor strength via peak torque values obtained from a series of two maximal voluntary isometric contractions in a Biodex dynamometer at a neutral ankle joint angle ($0°$) and at $20°$ knee flexion, which we normalized to body mass. We also used the dynamometer to quantify k_{AT} during passive, isokinetic ankle rotations from $20°$ plantarflexion to $30°$ dorsiflexion. Here, we record cine ultrasound images through a longitudinal cross section of the musculotendinous junction (MTJ) between the medial gastrocnemius and Achilles tendon (AT). We manually tracked time series of the MTJ position, which we transformed into a common marker coordinate system with a marker on the posterior calcaneus to estimate AT length change. We calculated AT force by dividing ankle torque during the passive rotations by a generalized moment arm length [2]. Finally, we calculated k_{AT} from the slope of the linear best fit relationship between AT force and tendon elongation between 20-80% of passive range. The same subjects also completed two treadmill walking trials. Specifically, in randomized order, participants walked for 2 minutes normally and again while responding to a series of treadmill-induced slip perturbations (200 ms duration, 6 m/s^2) applied randomly 5 times bilaterally at heel strike. We collected 3D motion capture from the trunk, pelvis, and legs and used scaled musculoskeletal models to estimate two balance outcomes - namely: (i) 3D whole-body angular momentum (WBAM), and (ii) anterior and lateral margins of stability (MoS).

Results & Discussion: Older adults had lesser k_{AT} during passive rotation than younger adults (4.33 ± 1.78 N/m vs. 6.12 ± 2.82 N/m, $p = 0.016$), but indistinguishable values of plantarflexor strength ($p=0.865$). We found no significant associations between k_{AT} and WBAM. However, across our study cohort, higher values of k_{AT} associated with lesser perturbation-induced changes in ML MoS ($r=-0.303$, $p=0.048$). Similarly across our study cohort, we found that lower plantarflexor strength associated with higher frontal ($r=-0.350$, $p=0.021$) and transverse ($r=-0.402$, $p=0.008$) WBAM range. Those latter correlations were also evident but moderately stronger in the older adult cohort alone. Finally, only in older adults, we found that lower plantarflexor strength associated with larger lateral MoS ($r=-0.521$, p=0.016).

Significance: This study points to potentially-modifiable factors (i.e., ankle strength and k_{AT}), that could help older adults mitigate walking instability following a perturbation.
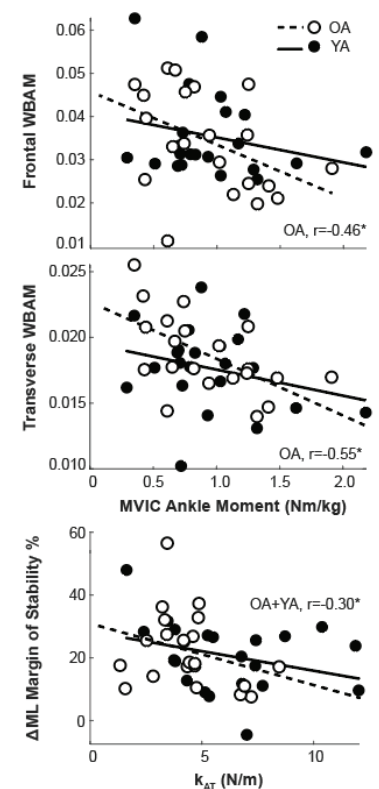
Figure 1. WBAM in the frontal (top) and transverse (middle) planes versus peak ankle torque for young adults in light grey (YA) and older adults in dark grey (OA). Mediolateral MoS (bottom) versus Achilles tendon stiffness (k_{AT}). Lines of best fit plotted for YA, OA, and Combined (black) with significant correlations (a=0.05) denoted by r-values.

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The interactive effects of biological tendon and ankle exoskeleton stiffnesses on walking metabolic cost

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Introduction: Passive-elastic ankle exoskeletons (EXOs) aim to improve walking economy by facilitating the storage and release of elastic energy during push off, which reduces biological ankle moments and requisite activation. EXO energy storage and return is governed by spring stiffness (k_{EXO}), which intuitively interacts with biological tendon stiffness (k_T) to influence overall device efficacy. Biological k_T has substantial effects on walking economy [1] and can change considerably with age [2]. Using experimental approaches alone, it can be challenging to document the interaction between k_T and k_{EXO}. Musculoskeletal modelling may streamline EXO tuning and personalization by identifying a candidate k_{EXO} that yields the most favorable interaction with k_T to reduce walking metabolic cost.

Our goal was to document the metabolic cost landscape across a physiological range of simulated k_T while walking at various k_{EXO} in a representative younger and older adult. We first hypothesized that ankle EXOs would reduce ankle extensor metabolic cost by decreasing requisite activation. We also hypothesized that ankle EXOs would elicit maximal reductions in ankle extensor metabolic cost at different k_T and k_{EXO} combinations for a representative younger versus older participant.

Methods: Two adults (*younger (YA):* 22 years, 1.84 m, 79.1 kg; *older (OA):* 70 years, 1.65 m, 61.1 kg) walked at 1.25 m/s in *Dephy ExoBoots* emulating a passive-elastic device with k_{EXO}s of 50, 150, and 200 Nm/rad. Participants received over 60 minutes of walking habituation across the various k_{EXO}s. We recorded whole body motions and forces as well as ground truth metabolic cost via indirect calorimetry. We measured k_T via isometric contractions and ultrasound of the Achilles tendon. We then customized scaled gait2392 OpenSim models for each participant to include a point mass with EXO inertial properties welded to each lateral tibia segment. We represented each EXO as an external torque applied bilaterally to the talus and tibia. We modified k_T in each model by simulating different tendon strains at maximum isometric force (ε_0) by performing simulations at participants measured ε_0 (YA: 3.4%; OA: 6.7%) and at 2, 4, 6, and 8% ε_0. We determined muscle-level metabolic costs using a bioenergetic model [3] and iterated simulations across each k_T and k_{EXO} combination to determine effects on ankle extensor and whole-body metabolic cost during walking.

Results & Discussion: YA and OA exhibited similar profiles of measured metabolic cost across k_{EXO} (Fig. 1A). Consistent with previous work [4], lower stiffnesses were most economical for both participants and increasing k_{EXO} raised walking metabolic cost. EXOs reduced measured metabolic cost to values below walking without the device in OA but not YA.

In silico, walking with increasing k_{EXO} reduced peak ankle dorsiflexion, moment, and activation for the ankle extensors. Walking with an EXO increased soleus metabolic cost for the YA (Fig. 1B) but decreased soleus metabolic cost for the OA (Fig. 1C). Increasing k_T (reducing ε_0) also decreased soleus metabolic cost across all k_{EXO}s. We found similar cost landscapes for the gastrocnemii.

Significance: Our simulations suggest that passive-elastic ankle EXOs may be more effective at reducing ankle extensor metabolic costs for older adults than for younger adults. One interpretation may be that younger adult muscle-tendon unit function and coordination during walking is likely to be sufficiently cost-optimized compared to older adults. One example of this is growing evidence that older adults exhibit lesser Achilles tendon stiffness than younger adults, which can disrupt ankle muscle mechanics.

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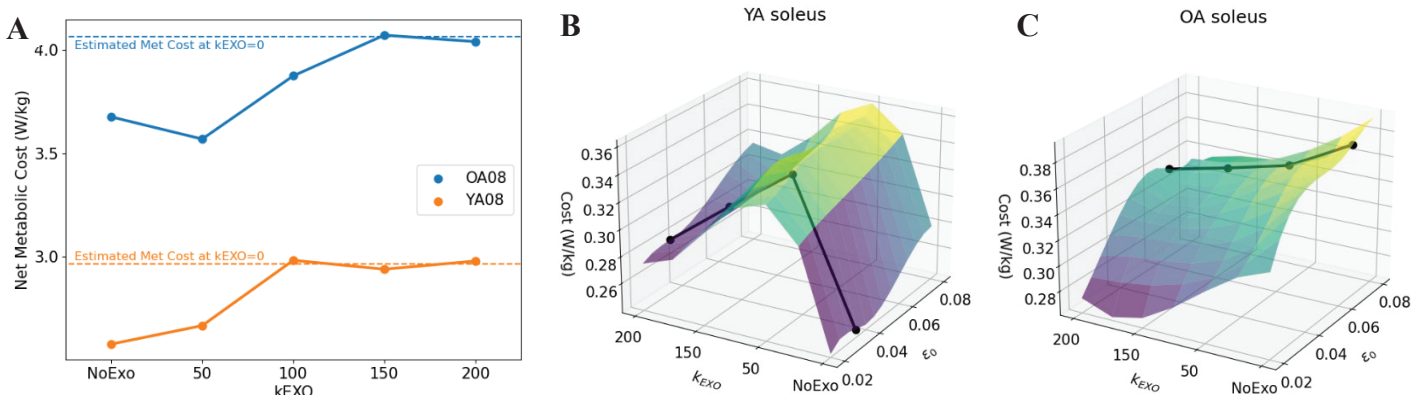
Figure 1: A) YA & OA displayed similar profiles of empirical metabolic cost across k_{EXO}, although OA walked with higher costs. B) YA soleus metabolic cost increased when wearing the EXO. C) OA soleus cost decreased when wearing the EXO. Black dots indicate cost at biological k_T.

AN IMPLANTABLE ACTUATOR FOR MUSCLE FORCE ASSISTANCE IN A BIPEDAL ANIMAL MODEL

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Introduction: Mobility, a key metric for quality of life, is often rehabilitated using basic aids (e.g. walkers). Wearable robotics, such as exoskeletons, offer a sophisticated solution, but often results in non-adherence [1] due to their impracticality, limiting their efficacy. Furthermore, externally-worn devices make precision and control challenging. Here, using a bipedal animal model (*Numida meleagris*), we present a concept for a fully implantable assistive limb actuator that can provide assistive torque at the ankle and reduce muscular demand at no additional energy cost. Rather than adopting an actuator that inputs mechanical power, we emulate the *in vivo* mechanics of leg muscles functioning isometrically that use tendons to store and release mechanical power [2]. This approach is similar to the externally-worn synthetic Achilles tendon used to reduce metabolic costs in humans [3]. Specifically, our aims are to develop a surgically-implanted actuator capable of: a) tight integration with biological tissues to replace one of the calf muscles and assist the remaining native calf muscles during gait; b) emulating native *in vivo* isometric loads and tendon energy storage/release; c) precise and rapid length and load adjustment (i.e. adaptable to shifting mechanical requirements) and clutching (force engagement/disengagement).

Methods: The basic control principle applied here consists of engaging an assistive actuator strut that replaces part of the native calf musculature during the stance phase, and subsequently disengaging the strut during the swing phase. Actuator design: a micro-DC motor paired with an encapsulating lead screw was used to change the length of the actuator body, while keeping the actuator volume low. The lead screw is designed to hold the actuator length static whilst under axial force applied by the biological tendon, without the need for actuator activation and energetic cost, while also providing rapid length changes during unloading (limb swing). The lead screw design was chosen with a nominal diameter of 8 mm and a thread pitch of 1.25 mm (Fig. 1). Implantation: implantation was developed on cadaver specimens (n = 3) and in survival surgery (n = 3). Animals were obtained as day-old animals from a commercial hatchery and cage reared to skeletal maturity. All experiments were approved by the PSU Institutional Animal Care and Use Committee. Surgeries were performed under standard aseptic and general anesthesia techniques (isoflurane inhalation and epinephrine/lidocaine injection for bleeding control). A blunt dissection was used to access the lateral aspect of the proximal tibia for the proximal actuator-to-bone attachment using a 2.2 mm micro corkscrew bone anchor with an integrated 2.0 FiberWire (Arthrex Inc.). The lateral gastrocnemius (LG) muscle was resected, creating functional deficit and freeing space for the actuator implant and room on the LG tendon for the distal attachment of the actuator. A Kessler suture technique was used to secure the distal actuator to the native LG tendon. *In situ* testing: A setup using cadaveric specimens was used to validate the mechanical feasibility of the micro-actuator. A stereotactic rig was designed to record forces and displacement prescribed to the biological tendon with and without the implantable actuator. The test bench consisted of a dual-mode force-length transducer (Aurora Scientific 310CLR lever, Aurora, ON, Canada; Fig. 1) mounted on linear rails and motion systems that allow the lever arm multiple degrees of freedom and adjustability. Lowman bone clamps were used to attach the femur to the rigid structure (Fig. 1). With the actuator inserted, the common Achilles tendon was severed distal of the ankle joint. The free tendon was attached to the Aurora lever arm using inextensible suture. The actuator was tested under ramp loading and tendon loading profiles simulating gait mechanics (OpenSim; [4]).
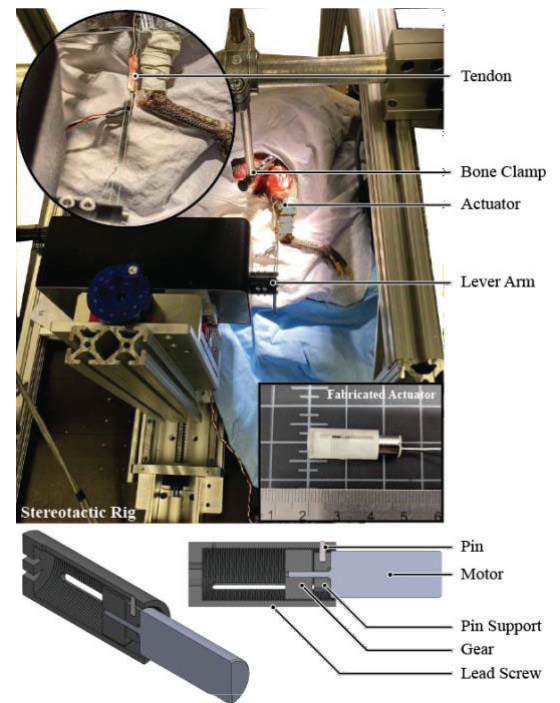

Figure 1: Characterization of micro-actuator within a cadaveric setup using a custom stereotactic rig.

Results & Discussion: We have developed a variable length actuator based on the concept of 'strutlike' biological muscle function (measuring ⌀9 x 30 mm) that can be fully implanted within the leg via a bone anchor and tendon fixation, replacing the lateral gastrocnemius muscle. The actuator is able to generate isometric force similar to the *in vivo* force of the native muscle during gait (~40N). The device has a stroke of 10 mm that operates up to 770 mm/s (77 stroke lengths/s; on an order of magnitude faster than biological muscle), capable of rapid contraction and elongation under low load. The stroke characteristics permits rapid clutching (disengaging when needed) and a tunable slack length to modulate the timing and level of assistive force during gait. Surgical viability has been established using survival surgeries, showing no signs of device rejection.

Significance: This work establishes the feasibility of an implantable assistive limb actuator using an animal model. Ongoing development aims at actuator miniaturization, sensor integration, and dynamic real-time actuator control in gait trials.

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TOO MUCH EXOSKELETON "ASSISTANCE" CAN DISRUPT USER BALANCE CORRECTION

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Introduction: Wearable robotics such as exoskeletons have the potential to enhance balance [1,2,3,4]. We previously showed that providing humans with ankle exoskeleton torque before muscle response in backward support surface standing balance perturbations could increase balance capacity. In large perturbations, 30 Nm of plantar-flexing exoskeleton torque provided to the user's ankles reduced center of mass (CoM) kinematics, improving balance capacity [1]. However, it remains unclear whether such exoskeleton assistance improved balance following smaller perturbations. As users will experience a variety of perturbation outside the laboratory, appropriately tuning assistance magnitude is important for designing a controller for balance-improving wearable robotics. In this study, we sought to evaluate whether the amount of torque given before human response, beginning around 150ms after perturbation onset, affects balance in smaller balance perturbations.

Because these perturbations typically elicit an ankle strategy for balance recovery, we hypothesized that, given the same perturbation and timing of assistance, higher levels of ankle assistance would improve balance more than lower levels of assistance. Thus, we predicted that greater torque delivered before the muscle response would reduce CoM acceleration, velocity, and displacement more than a moderate amount of torque.

Methods: Ten healthy adults (26 ± 2 yrs) wore ankle exoskeletons (Dephy ExoBoot, Dephy Inc.) and were instructed to maintain standing balance during backward support-surface perturbations (Fig 1A). An accelerometer on the exoskeleton detected perturbation onset in ~40ms, after which the exoskeleton was commanded to either provide 15Nm or 30Nm of plantarflexion torque (50ms rise time, followed by a decline to 0 Nm over 150ms) immediately (*15Nm Fast & 30Nm Fast*) or to provide no torque (*Off*). To prevent adaptation to the magnitude of exoskeleton assistance, conditions were randomized. In all perturbations, the support-surface traveled 15cm over 500ms. To confirm the delivery of exoskeleton torque, center of pressure (CoP) displacement was calculated from force plate data 100ms after perturbation onset. CoM kinematics were calculated from recorded motion capture and force plate motion data. To separate immediate effects of exoskeleton torque (<150ms) and the later response of the nervous system after assistance (>150ms), the integral of acceleration & velocity was computed in two time-bins: 0-200ms (initial phase) and 200-500ms (later phase) following perturbation onset. Within-subjects repeated measures ANOVA statistics were run on to evaluate differences between conditions.

Results & Discussion: Considering CoM kinematics as a metric of balance performance, early exoskeleton assistance improved balance during the initial phase of perturbation. CoP was shifted anteriorly $1.7cm \pm 0.8cm$ more in *15Nm* and $1.6 \pm 1.0cm$ in *30Nm* compared to *Off* (p<0.01), confirming that torque was delivered earlier than human response. The integral of CoM velocity reduced by 2.45 ± 0.2 mm in *15Nm* and by 2.4 ± 2mm in *30Nm* compared to *Off* (p<0.01 & p=0.03); was no difference between *15Nm* and *30Nm* (Fig1B &C).

However, in some individuals, 30Nm of assistance made their balance worse, as evidence in the late phase (Fig 1B right vs left pane). In the *30Nm* condition, six participants had 5.5 ± 0.8 mm lower integral of CoM velocity (Fig1B, left) while 4 participants had 8.3 ± 0.6 mm higher integral of CoM velocity in the only (Fig1B, right). Comparing the 30Nm to the 15Nm condition, 6 poeple had 0.90 ± 0.5 mm lower integrated velocity while 4 had 2.3 ± 0.4 mm higher integrated velocity (p<0.01, Fig1B&D).

Significance:
Wearable robotics can actually have a negative impact on balance recovery if they provide too much assistance, and that this effect does not apply similarly to everyone.

The appropriate level of balance-correcting torque in response to perturbed balance may differ from person to person, possibly due to an increased reliance on hip strategy. If inappropriately high, torque may end up becoming a balance perturbation itself to the user.

Figure 1: A) Experimental setup. B) CoM acceleration (top), velocity (mid), and displacement (bot) traces from two C) CoM travel in the initial and D) late phase for each condition. Each line represents a participant.

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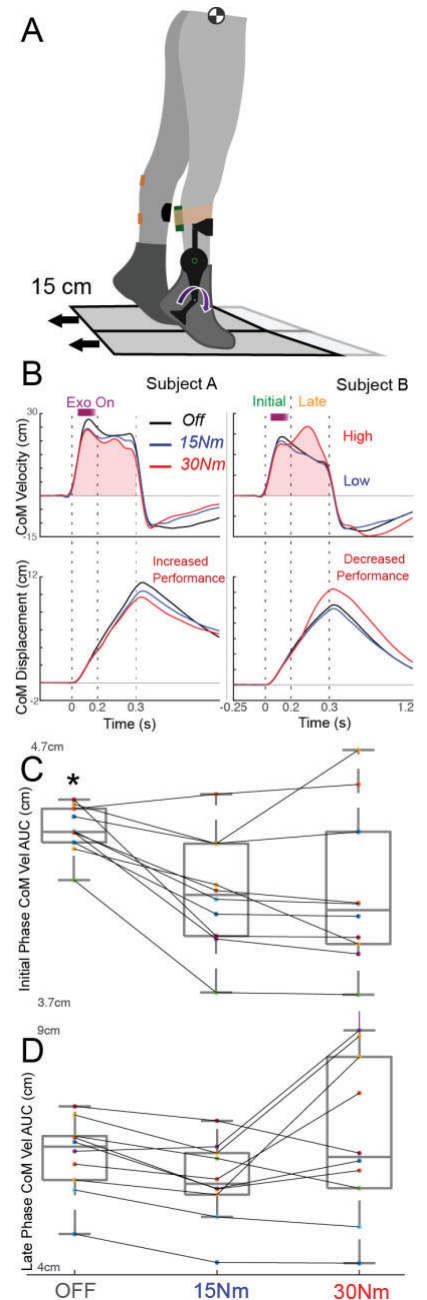
TRADING OFF STABILITY, AGILITY AND EFFICIENCY OF MOVEMENT ACROSS A BROAD RANGE OF MUSCLE-TENDON MORPHOLOGIES

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Introduction: Animals care about multiple locomotion objectives, and yet are stuck with one version of their bodies at any given time. This produces a conundrum, which physiological properties should animals pick? While answering this question is subjective, as it depends on the animal's specific needs and its ecological context, we can however start to discover what tradeoffs the animal does face in making this decision. Previous studies have compared different breeds of dogs [1] or different types of athletes [2] performing the same task to make conclusions about the relative pros and cons of one musculo-tendon parameter set over another. However, since this is correlational and not causal, and since it's difficult to experimentally change one parameter of an animal's morphology while keeping others constant, it makes it difficult for us to generate maps of physiological parameters to functional performance across multiple objectives/tasks. Therefore, in this study, we used a mathematical modeling approach. In particular, we asked – given a simple one-dimensional cyclic task such as hopping with your ankles – how does tendon stiffness affect the stability, agility and economy of the task independently and what are the tradeoffs animals face when picking one value of tendon stiffness over another over the objective space? Since previous work has shown that hopping energetics is optimal when the system is in resonance, we hypothesized that there exists a tendon stiffness that generates resonant hopping which is metabolically efficient – but difficult to deviate from and therefore less stable and agile. And that, as you move to higher or lower tendon stiffness from that optimal region, we would observe lower metabolic efficiency but higher stability and agility – an unavoidable fundamental trade-off.

Methods: We modified a previously published model of human hopping to include an aerial phase [3]. The model consisted of a mass in gravity that is cyclically actuated by a compliant muscle-tendon unit that is stimulated by a square wave of 2.5Hz and 10% duty factor. Once the model achieved stable hopping, we either (a) did nothing or (b) perturbed the height of the ground or (c) perturbed the timing of stimulation for a large range of tendon stiffnesses. We measured the metabolic efficiency of hopping without a perturbation (measured by mechanical work/metabolic work), the stability of the system over various changes to ground height (measured by average cumulative mechanical work 5 hops after the perturbation/work injected by the change in ground height) and agility of the system (measured by maximum positive and negative work done by changing stimulation pattern/mechanical work without a perturbation).

Results & Discussion: We found that biological tendon stiffness values (represented by the black vertical line in the figure) coincided with maximum metabolic efficiency, intermediate stability, lowest agility (in terms of generating positive work) and close to highest agility (in terms of generating negative work). Reducing tendon stiffness from that point, reduced metabolic efficiency, increased stability, increased positive agility and first increased and then decreased negative agility. Subsequently, increasing tendon stiffness from the biological value, drastically reduced efficiency, drastically reduced stability, increased positive agility, and drastically reduced negative agility. Thus, there lies a pareto optimal front going from low tendon stiffness values up to biological values, where humans may trade efficiency for higher positive agility and stability.

Significance: Our finding of pareto optimality is particularly interesting because previously most studies focused on evaluating populations and interventions based on a single metric. For example, it was believed that older adults having lower tendon stiffnesses than young adults is a maladaptive trait as it may reduce efficiency [4] and thus needs to be corrected by exercise or exoskeletons. However, if our results could extend to other tasks like running and walking, then we may find that older adults are simply trading metabolic efficiency to gain higher agility and stability – which is a reasonable assumption as falls are highly risky in terms of health outcomes for older adults. Thus, an intervention to alter their tendon stiffness or account for it may in fact hamper their objectives. Thus, future research needs to contend with the multitudes of animal objectives and how it relates to their underlying morphology.

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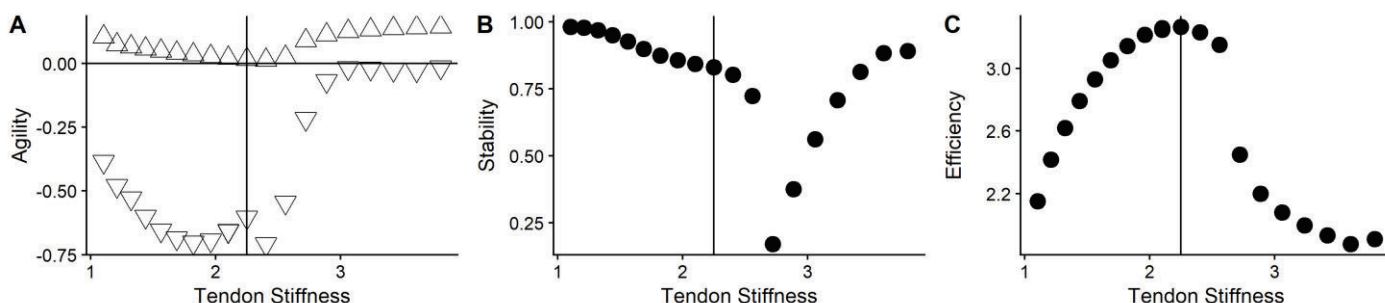
Figure 1: (A) Positive (triangle) and negative (downward facing triangle) agility, (B) Stability and (C) Efficiency of hopping across a broad range of tendon stiffness values. Note : Tendon stiffness on the X axis is normalized such that 2.25 relates to 180000N/m.

Prolonged Load Carriage During Walking Induces Fatigue and Redistributes Lower Limb Muscle Effort

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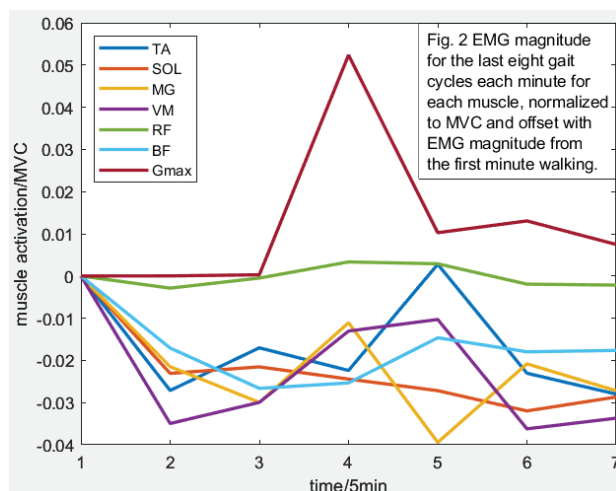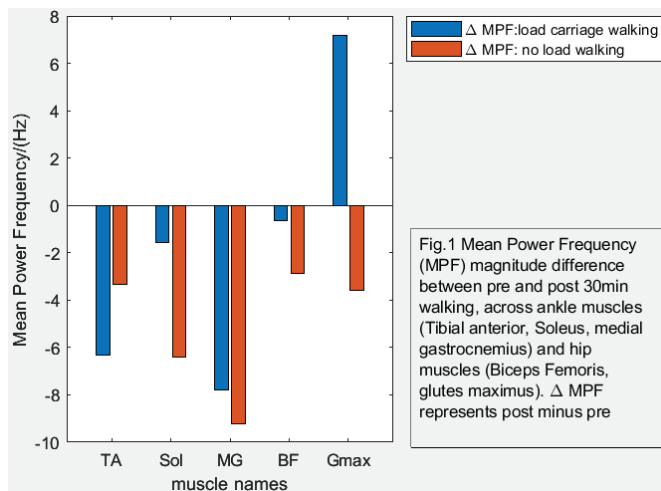[3]The School of Biological Sciences, Georgia Institute of Technology, Atlanta, GA, USA
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Introduction

Load carriage, muscle fatigue or a combination of both could increase falls risk because of changes in gait parameters that lead to balance challenges. For example, populations that engage in activity involving navigation through unstructured terrain while carrying body-borne load, such as soldiers, EMTs, and hikers, may be especially susceptible to falls because added mechanical demand could induce and neurological fatigue over time [1]. A previous study of muscle-level fatigue compared the max effort force and muscle activation magnitude before and after an acute bout of walking with 40% body weight and failed to establish significant changes in biomarkers of fatigue for muscles around the knee joint [1]. However, that study did not examine muscles around the ankle, knee and hip together during a prolonged walking trial, in conditions with vs. without load carriage. Here we begin to address this gap, setting up an experiment with the goal to characterize the *(re)distribution* of muscle fatigue across the lower-limb during load carriage over an ecologically-relevant timescale. We hypothesized that due to a higher relative demand for force and work to propel the body with added load, the distal muscles (e.g., ankle plantarflexors) would fatigue at a faster rate than proximal muscles (e.g., hip extensors) over a 30 min walking bout.

Methods

One healthy participant walked at 1.2 m/s on a split-belt treadmill mounted on a fixed-platform (CAREN, Motek Inc., Netherlands) for 31 minutes with (Session 1) and without (Session 2) a body-borne load. For the load carriage session, the participant donned a weighted vest with added mass equivalent to 20% of body weight. In the no-load session, the participant donned the weighted vest with no added mass. Lower-limb muscle electromyography (surface EMG), joint kinematics (motion capture), and ground reaction forces (force platforms) were collected every 5-minutes during each session. Gait cycles from each 1-minute recording were identified using the on/offset of the ground reaction force of a single leg. EMG data were bandpass filtered, offset by mean value, rectified, and smoothed prior to analysis. We calculated the mean power frequency and mean amplitude across the last 8 gait cycles of each minute long trial from EMG records taken from key muscles of the ankle (TA, Sol, MG) and hip (BF, Gmax).

Fig.1 Mean Power Frequency (MPF) magnitude difference between pre and post 30min walking, across ankle muscles (Tibial anterior, Soleus, medial gastrocnemius) and hip muscles (Biceps Femoris, glutes maximus). Δ MPF represents post minus pre

Fig. 2 EMG magnitude for the last eight gait cycles each minute for each muscle, normalized to MVC and offset with EMG magnitude from the first minute walking.

Results and Discussion

We observed a decrease in mean power frequency over the 30-minute walking bouts with (Session 1 – blue) and without body-borne load (Session 2 – orange) for *all* the muscles of interest except for gluteus maximus (Gmax) during the load carriage session (Fig. 1). This trend in our initial pilot data indicates that 30 minutes walking can induce muscle fatigue, even without added load [2]. In addition, during Session 1 with body-borne load, we found decreases in EMG magnitude for the ankle muscles (TA, Sol, MG) that was accompanied by increases in EMG magnitude for hip muscles (Gmax) (Fig. 2). This trend indicates a distal to proximal shift from using ankle muscles to using hip muscles that builds and persists as fatigue accumulates while walking with a body-borne load.

Significance

Characterizing the progression of fatigue over time during walking is a crucial first step that will help engineers develop assistive devices and develop training protocols that specifically target muscles that tire quickly. Our preliminary data suggests targeting the ankle muscles may be an effective means to mitigate local and global effects of muscle fatigue during load carriage. Future work will aim to extend these data and more comprehensively explore how joint-level neuromechanical effects of muscle fatigue impact dynamic balance.

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INTER-JOINT MONOSYNAPTIC FEEDBACK REDUCES SENSITIVITY TO PERTURBATION DIRECTION

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Introduction: Most muscle spindle pathways are relatively localized, but some are inter-joint and asymmetric [1]. Inter-joint asymmetric pathways may play a role in regulating whole limb properties [2]. The goal of this study was to understand how asymmetric muscle spindle feedback influences limb impedance and inter-joint coordination (limb mechanics). Intrinsic impedance (inertia + musculoskeletal +feedforward activation at yield) follows a high proximal and low distal gradient. The masses of the limb segments and muscles also follow a high proximal to low distal gradient, matching the impedance gradient. Despite these strong coupled gradients, joint excursions across the cat hindlimb appear equally distributed over many conditions, such as up and down slope walking [3]. Cats have excitatory spindle pathways from the hip to knee and knee to ankle [1], leading us to hypothesize that these feedback pathways *increase* whole limb impedance and inter-joint coordination and *decrease* sensitivity to the direction of limb force perturbations.

Methods: To evaluate how inter-joint, asymmetric spindle feedback influences whole limb impedance and inter-joint coordination, we used a two joint, two segment model in Simulink with parameters taken from feline biomechanical [4] and neurophysiological data [5], with asymmetric spindle feedback modelled as off-diagonal stiffness and damping terms. We applied a varying direction ($\phi = -20^\circ to +20^\circ$, [6]) sinusoidal endpoint force with an average value of 60N to the end of the distal segment. We fit 10 steady state cycles of the output to an impedance equation that includes apparent K, B, and M [7] and calculated the amplitude of the hip angle divided by the amplitude of the knee angle at steady state to evaluation proportionality of joint excursions.
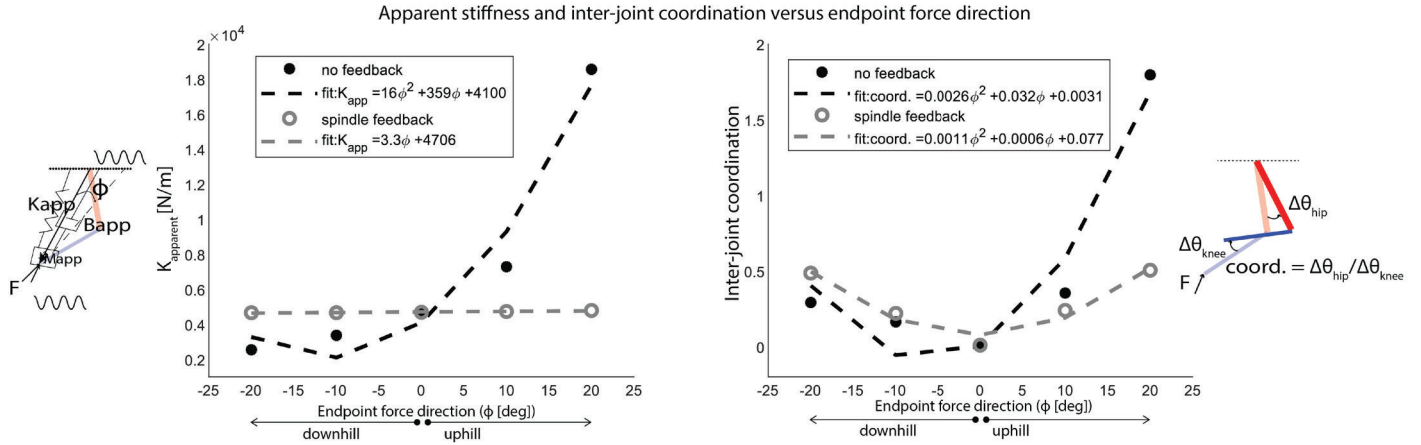

Figure 1: Apparent stiffness (left panel) and inter-joint coordination (right panel) versus endpoint force direction for no feedback (black) and spindle feedback (grey).

Results & Discussion: As shown in Figure 1, the apparent stiffness was higher with spindle feedback than with no feedback for endpoint force directions posterior to the hip (downhill). While the apparent stiffness changed significantly with respect to endpoint force direction (quadratic fit) without feedback, asymmetric spindle feedback caused the apparent stiffness to stay relatively constant (linear fit with high constant). Similarly, inter-joint coordination was higher with spindle feedback than with no feedback for endpoint force directions posterior to the hip (downhill). With no feedback, inter-joint coordination changed more significantly with respect to endpoint force direction than with spindle feedback (the first two quadratic fit parameters are higher without feedback). Therefore, in partial support of our hypothesis, asymmetric spindle feedback increases apparent stiffness and inter-joint coordination for endpoint force directions posterior to the hip and decreases sensitivity of both parameters to endpoint force direction.

Significance: Inter-joint coordination is important for maintaining muscles within their optimal (i.e., lengths best for force output) operating range of motion. Humans have extensive and bi-directional spindle pathways [8], so these pathways should be considered for diagnosis and treatment. Sensory pathways become disrupted after injury, and this leads to impaired inter-joint coordination (e.g., stroke [9]). Therefore, implementation of these feedback architectures in circuitry of assistive exoskeletons or FES support systems could better assist in recovery from injuries by providing adaptable forces and motions across many locomotor contexts.

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WHICH HIP MODEL BEST PREDICTS BIOLOGICAL TORQUES ACROSS LOCOMOTION MODES? A SIMULATION STUDY

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Introduction: Wearable assistive devices can augment people's mobility but lack a robust control strategy that can adapt to the various locomotion modes (speeds, slopes, and gaits) of everyday life. Human-in-the-loop optimization methods can find effective assistance at each mode, but these methods are time-consuming and exhausting for both the user and experimenter. Knowing that biological torque changes for each mode, we believe an analytical controller that could predict biological torque (BioTorque) across modes without user adjustment would address this issue. We simulated how well three optimized hip exoskeleton control models (impedance control (IMP), proportional myoelectric control (PMC), and muscle activity driven neuromuscular model-based control (NMM)) could mimic BioTorque across 26 different locomotion modes using only joint kinematics (angle and its derivative) and/or muscle activity (gluteus maximus and rectus femoris) as inputs (Fig. 1A). We hypothesized that the NMM would better predict BioTorque than IMP or PMC, as it leverages both kinematics and muscle activity.

Methods: We drove each model and comparison using previously recorded locomotion data from 5 participants including 4 speeds (1.25 m/s, 2 m/s walking, 2 m/s running, and 3.25 m/s running) across 7 slopes (±15° (except at 3.25 m/s running), ±10°, ±5°, and 0°). Each model was driven by the appropriate signals for that mode and the model torque output was compared against the measured, ground-truth BioTorque (Fig. 1C) using mean absolute error (MAE). The model parameters were tuned to minimize the MAE. To ensure we found the best tuning for each model, we compared three optimization algorithms (Surrogate, Bayesian, and CMAES) with enough iterations for each to converge within 5% MAE. We chose Surrogate, as it yielded the lowest error and the fastest convergence rate. IMP model took joint angle as input and then implemented a virtual spring and damper to generate an output torque (Fig. 1B). On the other hand, PMC model took measures EMG as input, amplified it (G_{EMG}) and concatenated it with a delay unit (Fig. 1B). Lastly, EMG-driven NMM-based controller [1] took in both joint angle and EMG as inputs, and implemented a Hill-type muscle-tendon model for each muscle, including a linear tendon spring in series with a contractile element representing muscle force and a non-linear spring in parallel (Fig. 1B). To calculate how well each model could mimic BioTorque at each trained mode, we split the data by gait cycles in a 3:1 training-to-validation ratio and measured the MAE only on the validation data. These results were averaged across participants and then the 26 modes for each model (Fig. 1D left). To determine how each model could predict BioTorque without retraining, we trained the model on one mode, measured MAE on the other 25 modes, and calculated the average per mode. We repeated this for all 26 modes and then averaged MAE across modes for each model (Fig. 1D right).

Results & Discussion: The IMP model best mimicked measured BioTorque, showing the lowest optimization (training) MAE at 0.17 Nm (Fig. 1D left) on average across participants and modes. PMC showed worst performance, with the highest MAE. Furthermore, IMP model, with only 6 tunable parameters, also outperformed the more complex NMM and PMC models in terms of adaptability, with an MAE of 0.25 Nm (Fig. 1D right). It seems incorporating measured gluteus maximus and rectus femoris muscle activity actually hinders BioTorque estimates, possibly due to their high variability and differences in peak timing compared to BioTorque. On the contrary, joint kinematics provide a less variable and more effective signal to drive analytical models.

Significance: IMP model, a simpler architecture with fewer parameters, can better predict BioTorque than more physiologically complex models. Leveraging this understanding will help design exoskeletons controllers focused on motion sensors, with controller based on joint kinematics to provide continuous assistance across modes.
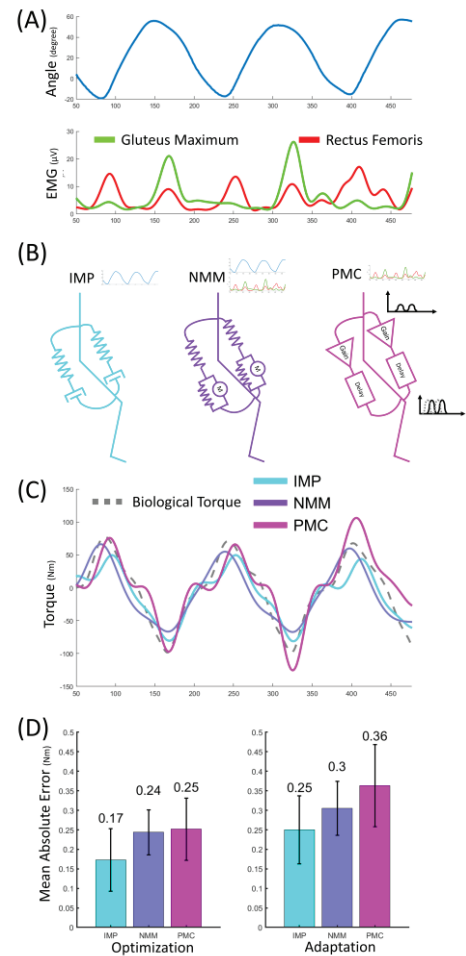

Figure 1: (A) Experimental angle and EMG data for model inputs. (B) Diagram of each model. The number of model parameters are 6 for IMP, 32 for NMM, and 4 for PMC. (C) Comparison between BioTorque and output torques of models. (D) MAE of optimization and adaptation.

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DO HUMANS USE MUSCLE ACTIVATION OR ENERGY COST TO SELECT WALKING SPEED? WHAT WE CAN LEARN FROM ANKLE EXOSKELETON INTERVENTIONS

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Introduction: Although the cause of slower walking with age is still unknown, exoskeletons may provide a viable solution for rapid increases in mobility for older adults. Walking speed declines as we age and has been correlated to reductions in quality of life and independence. Humans typically walk at speeds that minimize energy consumed per unit distance (cost of transport; COT) but this may not hold in older adults [1]. Changes in metabolic cost per unit muscle activation accompanying the distal-to-proximal shift in lower-limb joint power output with age [2] may dissociate the relationship between COT and lower-limb cumulative muscle activity per distance (CMAPD) in older adults. The resulting divergence in CMAPD vs. COT cost landscapes could help identify the underlying mechanism driving self-selected walking speed (SSWS) in humans. Exoskeletons provide a tool to modulate the COT [3] and muscle activity needed to walk at a given speed. Together, measuring changes in COT and CMAPD with exoskeleton assistance applied to younger and older adults could determine whether COT or CMAPD better correlates with changes in SSWS. We hypothesize that changes in the CMAPD rather than COT optimal speed will better correlate with changes in SSWS due to exoskeleton assistance.

Methods: We used Dephy ExoBoots to apply assistive ankle torque to 3 younger adults (YA) and measured their overground SSWS without the exoskeletons (NoExo) and with optimized assistance (Exo). To compare optimal COT and CMAPD speeds, we measured whole-body metabolic cost and muscle activity in 8 lower limb muscles (tibialis anterior, soleus, medial gastrocnemius, vastus medialis, rectus femoris, biceps femoris, gluteus maximus, gluteus medius) across 5 different speeds ($SSWS_{NoExo}$, $SSWS_{NoExo}$ +/- 33% & 67%, and $SSWS_{Exo}$). We fit a quadratic curve to the COT & CMAPD data across speeds for each Exo condition per each participant. Using the optimal speed (i.e., speed at minimum COT or CMAPD) for each curve, we plotted the percentage difference (Exo from NoExo) in COT and CMAPD vs. percentage difference in SSWS and fit a linear regression to the across participant data.

Results & Discussion: Despite tuning to optimize exoskeleton (Exo) assistance for increased SSWS, upon validation we found that Exos did not change SSWS (Fig. 1 A&C). Nevertheless, Exo assistance did lead to measurable changes in the speed for min COT and min CMAPD (Fig. 1, A-D). Optimal speeds for min COT and min CMAPD were slower than the associated SSWS, likely due to measurement location (treadmill for COT and CMAPD, overground for SSWS). Overall, Exo use decreased COT at optimal speeds, increased COT at the slowest and fastest speeds (Fig. 1A) and did not have a large effect on CMAPD across speeds (Fig. 1C). Changes in optimal COT speed were negatively correlated with changes in SSWS (Fig. 1B), while changes in CMAPD were positively correlated (Fig. 1D). Overall, CMAPD more strongly correlated with changes in SSWS than COT.

Significance: Exoskeletons can be used to understand more about human behaviour by driving changes in users' physiological response. Here we have shown that COT may not be the best predictor for SSWS compared to CMAPD. This may be intuitive, as the body has no way of directly measuring energy consumption, but does have sensory organs that correlate with muscle loading (e.g., spindles, Golgi tendons). Interventions that reduce relative muscle activation (i.e., making muscles 'stronger') could more directly affect walking speed selection. A follow-up study with older adults that includes hip Exo assistance will compare changes in SSWS across target joints.
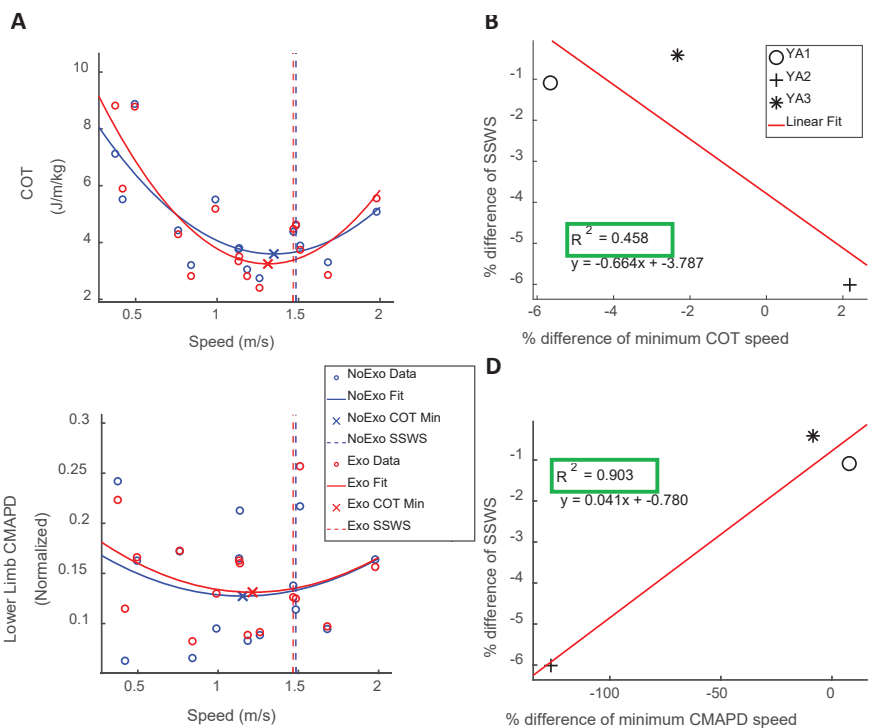

Figure 1: (A) COT vs Speed for all subjects (N = 3) for Exo (red) and NoExo (blue), each fit with a quadratic curve and its minimum marked. (B) Linear regression between percentage change between Exo and NoExo in optimal COT speed and SSWS per subject. (C) Lower Limb CMAPD vs Speed for all subjects (N = 3) for Exo (red) and NoExo (blue), each fit with a quadratic curve and its minimum marked. (D) Linear regression between percentage change between Exo and NoExo in optimal Lower Limb CMAPD speed and SSWS per subject.

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DEVELOPMENT OF A FOOT-ANKLE EXOSKELETON TO REDUCE THE METABOLIC COST OF DISSIPATIVE TERRAIN LOCOMOTION

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Figure 1: A picture of the foot-ankle exoskeleton (top), normalized metabolic rate for hard ground, sand, and augmented hopping (middle) and fascicle length relationship (l/l_0) (bottom) for the tested participant.

Introduction: Sand, gravel, and other surfaces consisting of granular media have more complex interactions with a human's walking gait than static surfaces (like a paved road). These surfaces often show non-linear and dissipative behavior, resulting in them being sensitive to applied forces, dynamically changing over time, and energy intensive to traverse. Currently, lower-limb exoskeletons are not designed or optimized for these surfaces. This gap in functionality was the focus of this study; the goal being to design an exoskeleton meant specifically to handle the challenges posed by sandy terrain, while reducing the energetic cost of movement.

Our previous work [1] has shown that much of the increase in metabolic cost of dissipative terrain locomotion is due to unfavorable muscle length operating points and higher muscle velocities. Additionally, *in-silico*, we have shown that a foot-ankle exoskeleton with optimized geometry and stiffness can eliminate the metabolic penalty associated with energy dissipation during locomotion on dissipative substrates. Now, we seek to translate these ideas into a bioinspired physical prototype that combines added area and added stiffness around the ankle joint, that we will used to comparatively investigate assisted and unassisted human hopping in sand. To this end, based on previous findings from modelling and human walking studies, we hypothesized that participants would (1) show a decrease in metabolic cost when performing a mechanically matched task with our exoskeleton on sand versus without and (2) operate at more favorable (longer) muscle lengths and lower fascicle velocities.

Methods: After obtaining informed consent, one participant (age: 22 years; height: 1.784m; mass: 88.8kg; resting metabolic rate 1.53 W/kg) was asked to fast overnight, and upon arriving in the morning, a 10-minute standing metabolic baseline was taken using an indirect calorimetry system. The participant was asked to don a pair of force sensing insoles and reflective makers were placed at the left and right posterior superior iliac spine to provide height biofeedback and validation through motion capture. A B-mode ultrasound probe was then placed on the skin superficial to each participant's right soleus, and the subject was then asked to perform 4 x 5-minute hopping trials. Trials were performed at 2.5Hz to matched height on hard ground, on sand, with only the exoskeleton foot on sand, and with the entire exoskeleton on sand. Terrain conditions were randomized, and metabolic expenditure was measured through the same means as the resting trial.
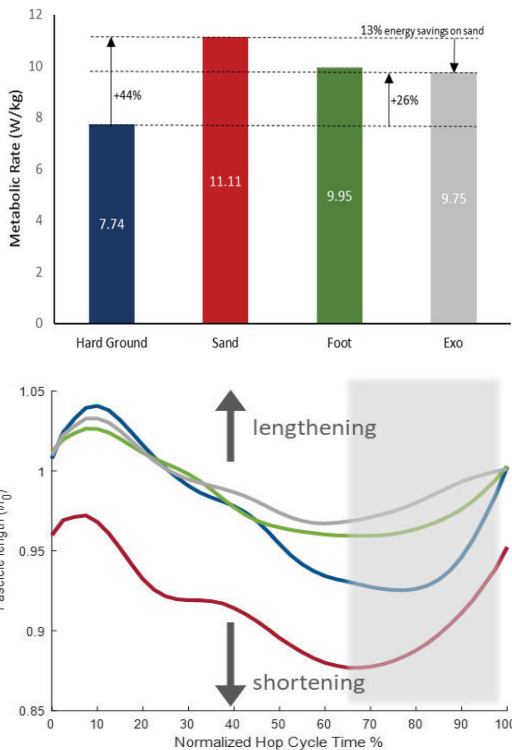
Results & Discussion: In agreement with our previous modelling and pilot results, we found a decrease in metabolic cost when performing a mechanically matched task with the exoskeleton on sand versus without. We found decreases in fascicle operating lengths on sand of when compared to mechanically comparable hopping on hard ground, with the fascicle lengths shown as a function of l_{CE}/l_0. Additionally, we found that the addition of the foot and the exoskeleton returned the operating length of the muscle to a similar region as that of hard ground. Unlike the terrain-length relationship, we found no appreciable changes in the fascicle shortening velocities. We expected to see a reduced shortening velocity due to the reduction in metabolic cost, and thus will continue to investigate this as the number of enrolled subjects increase. We found little difference between the hard ground and assisted conditions, prompting the investigation of additional MTU systems around the knee and hip.

Significance: This study presents the first metabolic-cost-reducing device designed for use over dissipative terrain. These results serve to further the scope of knowledge of bipedal locomotion and are a first step in the design of wearable devices that can mitigate energetic penalties associated with 'real-world' locomotion over dissipative terrain for applications in healthcare, agriculture, and beyond.

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LINKING THE TERRADYNAMICS, MECHANICS AND ENERGETICS OF DISSIPATIVE TERRAIN LOCOMOTION

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Introduction: The mechanics and energetics of human locomotion on deformable media have been studied for decades, with works examining the metabolic expenditure and joint level effects across various forms of locomotion. While these works have linked whole body mechanics to energetics, little has been done to understand the underlying causes associated with the increased cost of transport in humans at the joint, muscle and ground surface level [1,2]. Much of this has been attributed to the limitations of instrumentation and sensing. Even so, it has been shown [1-3] that energy expenditure is not linearly scalable to actual mechanical work, with potential causation at the muscle level, however, there have been no human studies that allows for direct inverse dynamics from human locomotion over these substrates that directly link these mechanics to muscular function.

Our study presents a first step toward linking energetics to joint level mechanics, using a customized in-lab apparatus to perform the first comprehensive analysis of the joint-level inverse dynamic relationships for human locomotion on sand. The goal of this study was to understand why, at the joint level, it is more metabolically costly to locomote on sand, when compared to a similar task on hard ground. To this end, we hypothesized that participants would (1) show an increase in metabolic cost when performing a mechanically comparable task on sand versus hard ground, (2) Show increasing powers at across the joints of the leg, and (3) show that the increase in mechanical changes are not sufficient to account for metabolic rate differences.

Methods:

After obtaining informed consent, eight volunteers (average ± standard deviation; age: 23.63 ± 2.67 years; height: 1.78 ± 0.05 m; mass: 81.5 ± 10.1 kg; resting metabolic rate 1.48 ± 0.14 W/kg) were asked to fast overnight, and a 10-minute standing metabolic baseline was taken using an indirect calorimetry system. Participants were asked to don a pair of force sensing insoles and were prepared by administering a full lower body marker set. Participants were then asked to perform 2 x 5-minute hopping trials at 2.5Hz to matched height using biofeedback from the insoles and reflective markers on hard ground and on sand. Terrain conditions were randomized, and metabolic expenditure was measured through the same means as the resting trial.

Results & Discussion:

We found an increase in $\overline{P}_{met}$ of approximately 22% when comparing the mechanically matched hard ground and sand conditions. When the $\overline{P}_{mech}$ of the entire leg is considered, some interesting trends emerge. We found the total leg
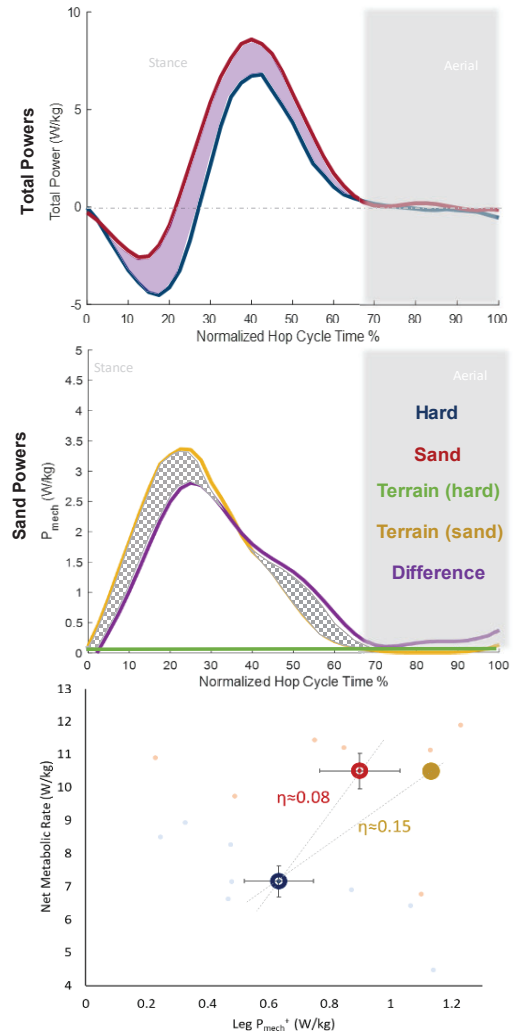

Figure 1: (Top) Power over the cycle for the leg on hard ground and sand, (middle) difference in powers over the cycle between the leg on hard ground as well as the actual terrains, (bottom) efficiency of P^+_{mech} of the leg from hard ground to sand.

power developed over hard ground to be $\overline{P}_{mech\,hard}^+ = 0.6399$ W/kg increasing on sand to $\overline{P}_{mech\,sand}^+ = 0.89$ W/kg, leading to an increased mechanical work input of $\Delta\overline{P}_{mech\,leg}^+ = 0.253$ W/kg. Comparatively, the power lost to the sandy terrain was given as $\overline{P}_{mech\,sand}^+ = 0.302$ W/kg over the cycle. This leaves a discrepancy of $\overline{P}_{mech\,sand}^+ - \Delta\overline{P}_{mech\,leg} = 0.05$. We hypothesize that these discrepancies were due to work done at joints not covered by our marker set, the midfoot. As such, investigation at the full body level set is needed to further determine the magnitude contribution of the additional joints. Finally, we found that the efficiency $\eta \approx 0.08$, lower than the expected efficiency of positive work, 0.25. When accounting for positive work contributed by other systems, η increases to only 0.15, assuming $\overline{P}_{mech\,total}^+$ to be equivalent to the total mechanical power performed on the sand over a cycle. This highlights inefficiencies in the lower limb muscles, and prompts deeper investigation into the muscle level effects of dissipative terrain locomotion.

Significance: We have presented a novel joint level energetic analysis that links the whole-body metabolic energy cost of hopping in sand, to the work done by the primary movers of the leg, and the work done on the surface itself. These results serve to further the scope of knowledge of bipedal locomotion in dissipate substrates and can further inform future studies at the muscle level, as well as the design of wearable devices that allows users to go further and farther than ever before.

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OLDER ADULTS BENEFIT MORE THAN YOUNGER FROM ACTIVE, BUT NOT PASSIVE, ANKLE EXOSKELETONS

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Introduction: Older adults consume metabolic energy faster during walking than young adults. Age-related changes in muscle-tendon units (MT) spanning the ankle are thought to cause reduced muscle power and force economy [1]. For example, due to reduced Achilles tendon stiffness (k_t), older adults walk using shorter, less economical muscle operating lengths and higher muscle activations than younger adults [2]. Previous literature in younger adults has shown passive ankle exoskeletons can reduce metabolic cost by altering the ankle muscle-tendon dynamics [3]. Additionally, active exoskeletons have shown even larger reductions in metabolic cost in real-world settings [4]. The purpose of this study was to compare passive and active ankle exoskeleton assistance strategies across age. Since older adults exhibit lower baseline economy, we predicted ankle exoskeleton assistance would result in larger metabolic benefits for older versus younger adults regardless of assistance strategy (i.e., passive vs active).

Methods: Seven young (5 M, 2F, 21 ± 2.2 yrs) and three older adults (3F, 73 ± 7 yrs) have thus far walked on a treadmill at 1.25 m/s. We programmed commercially-available portable ankle exoskeletons (Dephy Inc, Maynard, MA) to deliver spring [3] or motor-like assistance [5]. For each participant, we compared 6 exoskeleton assistance conditions: no exoskeleton, exoskeleton with zero torque (i.e., added mass), spring-like assistance (k_{exo} = 70 Nm/rad), and motor-like assistance at low, medium, and high magnitudes (T_{peak} = 10, 20, 30 Nm) (Fig. 1 inset). We measured $\dot{V}O_2$ and $\dot{V}CO_2$ using a portable indirect calorimetry system (COSMED, Italy) and converted to metabolic power using Brockway's equation.

Results & Discussion: Consistent with previous literature, we found ankle exoskeletons that applied both spring-like (~5% reduction) and motor-like (~7-18% reduction) assistance reduced metabolic cost for younger adults when compared to zero torque (Fig. 1a). In partial support of our hypothesis, older adults *only* benefited from motor-like assistance (~12-24% reduction) (Fig. 1b, blue). Surprisingly, metabolic cost increased (~5%) with spring-like assistance for older adults (Fig. 1b, pink). For motor-like assistance, older adults exhibited, on average, 5% larger metabolic benefit than young adults.

A potential mechanism driving larger improvements from motor-like assistance could reside at the muscle level [7], by shifting muscle dynamics to more economical operating points (i.e., longer lengths, slower velocities). Additionally, these results are from 41 (spring-like) – 57 (motor-like, 19 min at low, medium and high) min of exoskeleton assistance, and literature suggests full benefits take up to 218 min [8]. We believe the spring-like assistance was unable to provide a metabolic benefit for older adults because the k_{exo} needs to be optimized to their physiology. We would predict that a stiffer spring would be needed to offset known reductions in biological k_t in older adults.

Significance: Given the clear performance gap between motor-like over spring-like assistance, our results should motivate engineering developers to focus on creating powered solutions in leaner form factors at more equitable price points. This is particularly important when considering adoption of exoskeletons as assistive technology for people with mobility challenges.

Figure 1: The absolute change (left axis) and percent change (right axis) in net metabolic power (mean ± SE) from the exoskeleton zero torque condition for young (a) and older (b) adults. The inset plot is average exoskeleton torque patterns from one participant. Change in net metabolic power young: -0.503, 0, -0.110, -0.289, -0.668, -0.594 W/kg; older: -0.942, 0, 0.170, -0.534, -0.779, -1.105 W/kg respectively. Percent change in net metabolic power young: -13.437, 0, -2.925, -7.719, -17.856, -15.860 %; older: -20.305, 0, 3.669, -11.516, -16.794, -23.832 % respectively. This means motor-like but not spring-like ankle exoskeleton assistance was more metabolically beneficial for older adults.

Age-related differences in performance suggest a person's physiological properties may need to be considered when prescribing exoskeleton assistance. Future work will explore which physiological properties (e.g., k_t, soleus force-length-velocity properties) are most closely related to our observed age-related differences in metabolic benefit due to ankle exoskeleton assistance.

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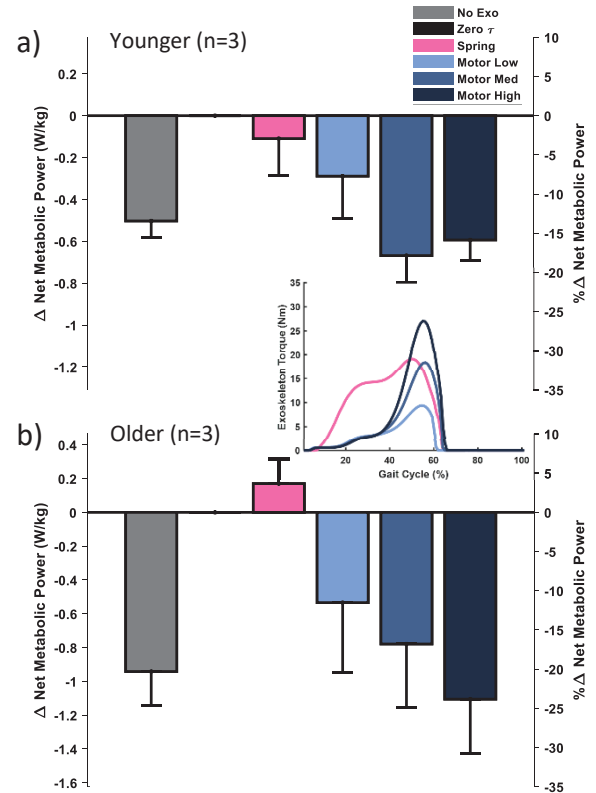
IMPACTS OF FATIGUE ON NEUROMECHANICAL STRATEGIES IN SINGLE-LEG JUMPING

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Introduction: Fatigue has a profound influence on neuromuscular and biomechanical properties in humans, particularly regarding the risk of injury around joints [1]. The behavior of joints and their corresponding muscles can be altered following physical exertion, via factors such as muscle activation and coactivation, force production capacity, motor control and movement patterns, and kinetics and kinematics, among many other properties [1]. Although fatigue is known to ultimately reduce force production capacity of the affected muscles, humans can maintain their performance level over long periods through compensatory strategies; multi-joint motor tasks use a range of neuromechanical strategies to prolong an activity at a given performance output [2]. This study aims to characterize changes in these strategies within the context of single-legged jumps (SLJs) following fatigue; this task is common across many sports and is often associated with injuries at the knee joint [3]. By investigating the subtle changes of measurable biomarkers before and after fatiguing the lower limbs, we hypothesized that humans rely less on joints and muscles that have been fatigued when performing explosive tasks involving power production and absorption, such as jumping. Therefore, we predict that the joints associated with muscles that experience a greater degree of fatigue contribute less to the net mechanical work of the task, and that less fatigued neuromuscular components act synergistically to compensate via a redistribution of energy within and between joints in an attempt to preserve net mechanical work done on the center of mass.

Methods: We collected preliminary data from one participant (30-year-old male, height =185.9cm, mass =111.1kg), with the goal of fatiguing to a point just before task failure, which may provide more ecologically relevant results. The participant performed three maximal SLJs using his dominant leg on a force platform. The maximum vertical displacement of these three jumps was recorded and used to calculate the target height for subsequent SLJs (75% of maximum height). After sufficient rest, the participant was tasked with a fatigue protocol in which he squatted repeatedly to a knee flexion angle of 90 degrees, with the goal of inducing targeted fatigue of the major muscle groups in the leg. Squatting occurred at a pace of 50 bpm until one of two possible criteria was met: the participant squatted for three consecutive minutes or failed to complete two successive squat cycles to the target flexion angle on beat. At this point, he attempted three maximal SLJs to the target height. If the target was reached in any of these jumps, the fatigue protocol was repeated until all three SLJs failed to reach the target height. Joint kinetics and kinematics were compared between the pre-fatigue and the penultimate cycle of the post-fatigue conditions.

Results & Discussion: We observed reduced net power production and absorption, and a redistribution of joint contributions to total mechanical work following fatigue (Fig. 1). The ankle was the principal power producer in the lift-off stage of the jump in both conditions. However, positive contributions of the knee were reduced by 15% post-fatigue. The positive power contributions of the ankle and hip increased by 13% and 2%, respectively, to counterbalance changes at the knee. In the landing stage, power absorption at the knee was reduced post-fatigue and the negative work contribution by the knee decreased by 14%. This reduction was offset by an increase in negative work performed at the hip and ankle by 4% and 10%, respectively. We speculate from these results that the knee experienced greater fatigue than the ankle or hip, resulting in the ankle's increased contribution to propulsion after neuromuscular fatigue. Similarly, the reduced contribution of the knee to the total negative work during landing points to greater fatiguability of knee extensor muscles. Moreover, the subject adopted a jumping control strategy that spared the knee through greater power absorption at the hip and ankle. The joint-level compensatory trends observed prior to SLJ performance decline were also observed after fatigue to task failure was established, highlighting the benefits of a progression-to-fatigue protocol. Previous research has found that humans exploit kinetic redundancies to maintain functional motor outputs, so the variance in joint work contributions seen in our findings provide motivation to further investigate the underlying mechanisms of motor abundance in SLJs and similar explosive tasks [4].
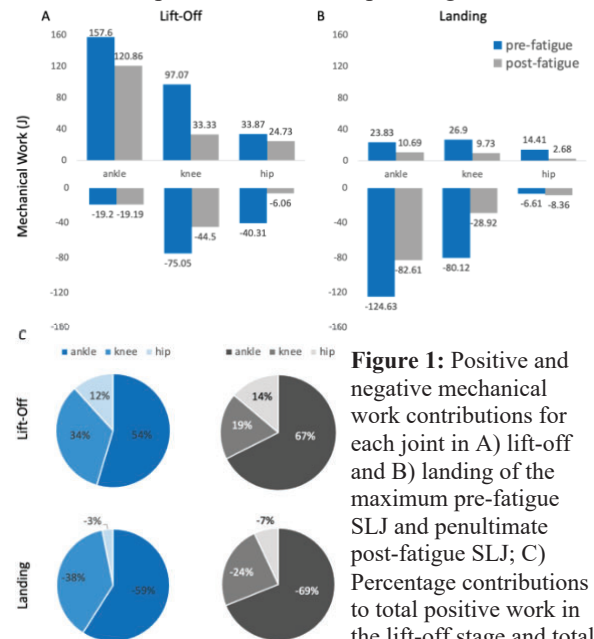

Figure 1: Positive and negative mechanical work contributions for each joint in A) lift-off and B) landing of the maximum pre-fatigue SLJ and penultimate post-fatigue SLJ; C) Percentage contributions to total positive work in the lift-off stage and total negative work in the landing stage.

Significance: These preliminary results highlight the presence of built-in dynamic control strategies to maintain performance outputs following neuromuscular fatigue. The involuntary modifications to joint coordination we observed allow performance to be maintained despite fatiguing conditions. The broader application of these results suggests the utility of identifying early biomarkers of fatigue to better understand the state of lower limb components during continuous, strenuous activities well before the point of injury. The characterization of such biomarkers in tasks such as single leg jumps will provide a framework for improved prediction and mitigation of fatigue-related injuries.

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RELATIVELY SMALL CHANGES IN FOOT-ANKLE MECHANICS OVER MONTHS ALTER MUSCULOSKELETAL STRUCTURE

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Introduction: Form follows function. Under lengthening conditions, skeletal muscle fascicles lengths increase by adding sarcomeres in series [1]. Following concentric or eccentric exercise, skeletal muscles hypertrophy[2]. On the other hand, muscle atrophy has been observed in chronic unloading and disuse studies. Tendons also adapt to their mechanical environments and are sensitive to several factors; strain magnitude, cycle duration, cycle frequency and intervention duration. Increases in tendon stiffness and cross-sectional area have been observed in exercise studies and tendon atrophy has been observed in chronic unloading and disuse studies. Most muscle and tendon adaptation knowledge comes from exercise intervention studies that include low cycle numbers and large strains. Literature suggests tendon is not likely to remodel unless interventions are above ~4.5-6.5% strain[3]. However, cross sectional studies of habitual high wearers report sizable increases in kt decreases in muscle length. This indicates that even with small changes in loading (i.e. low strain), muscle-tendon remodeling may occur if the number of cycles is plentiful.

The purpose of this study was to determine how relatively small changes in foot ankle mechanics remodels the underlying MT structure following relatively large number locomotion cycles. We altered foot ankle mechanics in daily activities over a 14-week intervention via modified footwear and measured leg MT structural properties pre and post intervention. We hypothesized that footwear that increased/decreased muscle operating lengths during daily use would lead to muscles that were relatively longer/shorter after intervention. We also hypothesized that devices that decreased/increased force on the calf MT over a large number of cycles would lead to tendons that were less/more stiff after intervention.

Methods: We recruited n=6 in the high toed footwear group and n=8 in the high heeled footwear group. Participants wore their experimental shoes during their daily activities for 12-16 weeks. We measured cross sectional area of the distal Achilles tendon (AT) using ultrasound. We used a dynomometer, ultrasound, and EMG to collect tendon and soleus muscle force-length relationships. AT stiffness is taken from the slope of the AT FL curve from 50-100% of MVC.

Results: During walking in daily life, our intervention increased fascicle operating length in the high toes and decreased fascicle operating length in the high heels [Abstract 146952]. Peak muscle tendon (MT) force increased in high toes by 21% and decreased in high heels by 8% [Abstract 146952]. Overall, tendon stiffness did not change in either the high toes ($p=0.456$) or the high heels ($p=0.449$) groups (Fig. 1A). AT CSA did not change in the high toes group ($p=0.133$), while high heels decreased AT CSA by 10% pre vs. post intervention ($p<0.001$) (Fig. 1B).

Figure 1: A) Achilles tendon stiffness (k_{AT}) pre and post intervention, n=4 toe group, n=8 heel group. **B)** Achilles tendon cross-sectional area (CSA) pre and post intervention, n=6 toe group, n= 8 heel group. **C)** Normalized soleus force-length relationships pre (light) and post (dark) intervention, n=3 toe group, n=2 heel group

Preliminary analyses suggest that shifts in soleus l_0 of the muscle for both high toes and high heels (Fig. 1C). Participants varied in the number of daily steps they look in their experiment shoes. In high heels, participants took avg ± sd: 1442 ± 1257 steps/day; range: 0 to 3704 steps/day. In high toes participants took avg ± sd: 3768± 1626 steps/day; range: 1179 to 5998 steps/day. For every 1000 steps/day participants took in high heels their k_{AT} increased 7-8% ($\beta= 19$ kN/m; $p=0.008$) and their CSA increased 5% ($\beta= 3$ mm^2 ; $p<0.001$). For every 1000 steps/day participants took in high toes their k_{AT} increased 7% ($\beta= 22$ kN/m; $p=0.019$) and their CSA did not significantly change ($p=0.535$).

Discussion: Although the high heel and high toe interventions changed triceps-surae MT force in opposite directions, AT stiffness increased in both groups when accounting for avg. steps/day in the intervention shoes. Further investigation of muscle-tendon dynamics and joint mechanics may reveal how tendon loading in the high heels led to increased k_{AT} and CSA despite reducing MT force.

Significance: Results from this study indicate that small changes in foot-ankle mechanics can result in changes in musculoskeletal structure with a high number of loading cycles. This study can increase our knowledge on musculoskeletal adaptation outside of exercise training and disuse protocols. It can also inform the wearable device field on how using devices that change foot-ankle mechanics over long-time scales can impact users.

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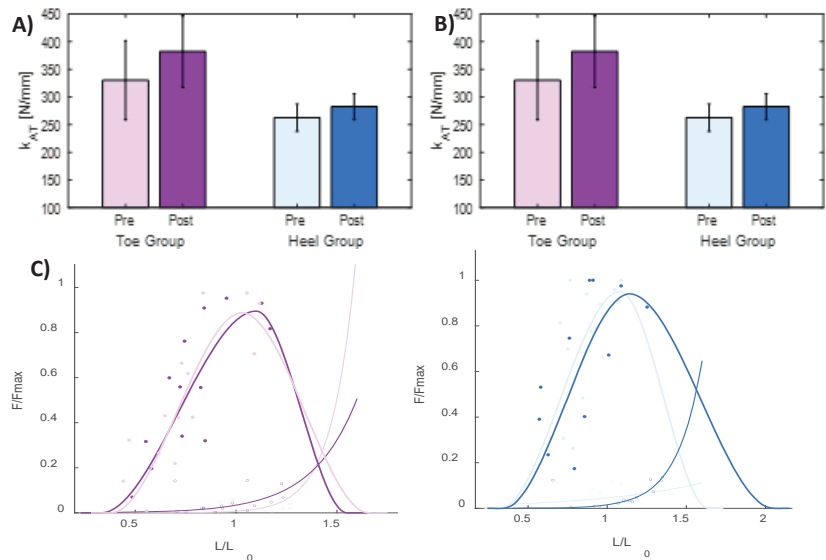
IMPACT OF TRICEPS-SURAE OPERATING LENGTHS ON WHOLE BODY METABOLIC COST

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Introduction: Lower-limb exoskeletons have been shown to reduce metabolic cost of walking by reducing muscle-tendon (MT) force [1,2]. Even passive ankle exoskeletons that do not add net mechanical energy to the user have shown energy cost reductions [3]. These devices may reduce costs by altering MT dynamics of the largest contributor to ankle push off, the triceps-surae [4]. Previously, we showed that during isometric ankle plantarflexion, decreasing soleus fascicle operating length by 17%, increased metabolic cost by 208% [5]. These isolated muscle contractions indicate that increasing the force potential of calf muscle reduces the metabolic cost of producing force by reducing the volume of muscle activated to perform a task [6].

The purpose of this study was to determine the impact of shifting triceps-surae operating lengths, during walking, on whole body metabolic cost. We used modified footwear, to systematically shift incur more plantar-flexing and more dorsi-flexing via raised heels and raised toes, respectively. We hypothesized that force on the MT would increase in high heels and decrease in high toes due to shifts in COP [7]. We hypothesized that walking in raised heels would decrease triceps-surae operating lengths, leading to an increase in whole body metabolic cost compared to walking in flat shoes. Similarly, we hypothesized the raised toes would increase triceps-surae operating lengths, leading to a decrease in whole body metabolic cost compared to flat shoes.

Methods: We recruited n=7 in high toe group and n=8 in high heel group. We instrumented participants with B-mode ultrasound over the medial gastrocnemius (MG) and soleus (SOL) muscle belly, a full body 3D motion capture marker set, and a metabolic mask. Participants walked for 5-minutes at 0.5, 0.9, 1.3, and 1.7 m/s in both experimental and mass matched flat shoes on a dual-belt Instrumented treadmill. We randomized trial order.

Results & Discussion: Both high heels and high toes increased the metabolic cost of walking compared to mass matched flat shoes (Fig. 1A). As expected, heels decreased SOL and MG operating lengths and toes increased operating lengths (Fig. 1B). High toes increased calf MT peak forces by 8%, while high heels decreased forces by 21% (Fig. 1C).

As expected, raised heel shoes increased metabolic operating cost, and decreased triceps-surae operating lengths. Even though high heels decreased force demand on the calf MT, the decrease in force potential via less optimal operating lengths may drive the metabolic cost increases [5]. Contrary to our hypothesis, raised toes also increased metabolic cost of walking despite increasing triceps-surae force potential via longer operating lengths. Increased force demand may contribute to the overall raised metabolic cost.

These results highlight tradeoffs in force demand and force potential at the ankle. Further analysis of more proximal muscles and joints can provide insight on how changes in triceps-surae force potential impact whole body energy expenditure during walking.

Significance: Metabolic cost reduction is a key outcome measure for lower limb wearables. Increased understanding on how triceps-surae MT dynamics impact whole body energy cost can inform the wearables field on how to build better, more effective devices. Furthermore, insights on how force demand and force potential impact energy expenditure can guide device design, control, and interventions.

Figure 1: A) Metabolic cost of transport curves in toe shoes vs flat shoes (n=7) and heel shoes vs flat shoes (n=8)
B) Soleus and gastrocnemius fascicle operating lengths during walking in flat and modified shoes at 1.3 m/s, for toe group (n=5) and heel group (n=5)
C) Triceps-surae MT force in flat and experimental shoes at 1.3 m/s, for toe group (n=5) and heel group (n=5)

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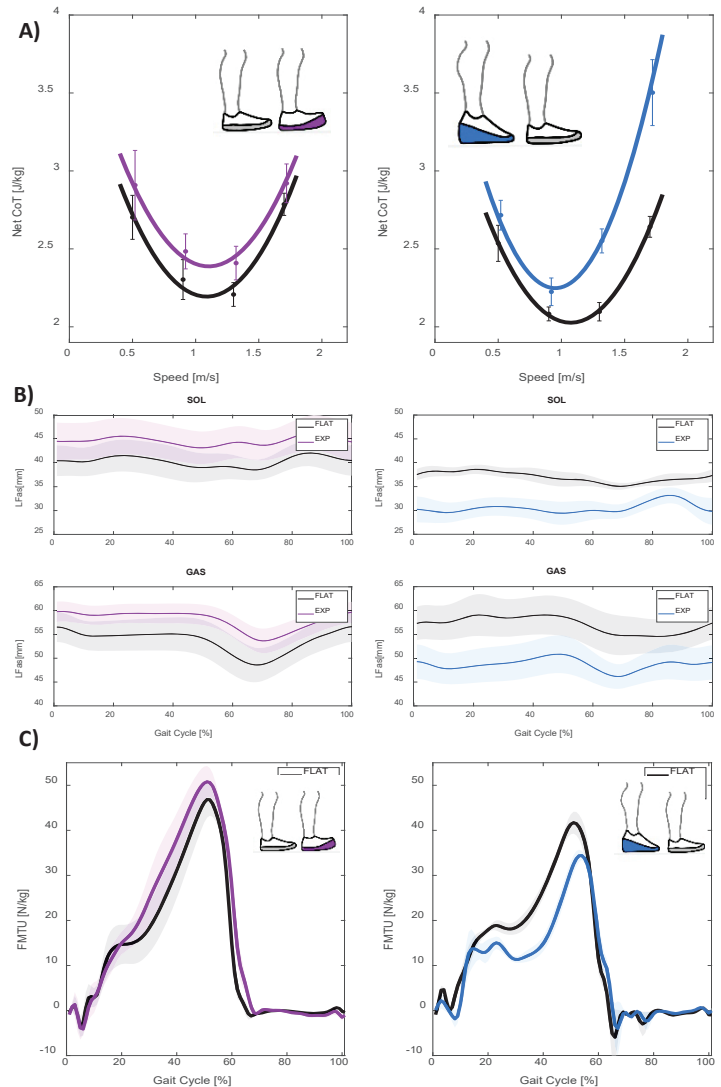
ANKLE EXOSKELETONS MITIGATE CALF MUSCLE FATIGUE OVER 30 MINUTE WALKING BOUTS

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Introduction

Balance assistance for clinical populations or for older adults is becoming increasingly necessary to support rehabilitation and prevent injury. To this end, walking balance control, fatigue analysis on a metabolic and muscular level, and exoskeletal device assistance have all garnered individual attention. Muscle fatigue correlates with reduced balance integrity – likely due to reduced force responsiveness on the muscular level [1]. Fortunately, exoskeletal assistance has been shown to decrease walking metabolic cost [3] and, at least with passive assistance under stationary loading tasks, mitigate fatigue [2]. Furthermore, it has been shown that proper ankle exoskeletal control can improve balance during walking [4]. Yet, the mechanisms by which exoskeletal assistance influences fatigue, and thereby fatigue-induced imbalance, remains unclear. We hypothesize that ankle exoskeletons assistance will offload force required for locomotion from the calf muscles to the exoskeleton motors, attenuate the onset of muscle fatigue, and result in reduced measures for biomarkers of fatigue in ankle plantar flexors.

Methods

Two participants (18-20 years) walked for a 30-minute period on a dual-belt instrumented treadmill platform (CAREN, Motek) on two days. A full body reflective marker set was applied to participants for motion capture, allowing for calculation of joint kinematics and kinetics. We applied surface electrodes to capture electromyographic (EMG) signals from the muscles depicted in Figure 1A. In session one the participant wore Dephy EB60 ankle exoskeletons, utilizing the optimized torque assistance profile controller developed Zhang et al. [3] with a 20 Nm peak torque. In session two the participant completed the same 30-minute walking bout without an exoskeleton. The order of the sessions was randomized between participants. We used two metrics to catalog evidence for muscle fatigue – namely, (i) mean power frequency (MPF) from each EMG signal and (ii) the ratio of integrated joint moment per integrated unit muscle activation for the muscles acting on that joint.

Results and Discussion

We have thus far found that ankle moment per unit soleus (SOL) activation and per unit medial gastrocnemius (MG) activation is greater with exoskeleton assistance than without, indicating reduced calf muscle fatigue with assistance (Fig. 1B). This result is consistent with the difference in MPF of the SOL and MG from minute 1 to minute 30, where exoskeleton assistance results in a higher MPF at the end of the walking bout compared to without assistance (Fig. 1A). Interestingly, we are finding that hip moment per unit gluteus maximus (GMAX) activation and per unit rectus femoris (RF) activation is less with than without exoskeleton assistance (Fig. 1B), indicating greater fatigue in the hip flexors and extensors with exoskeleton assistance. This result is consistent with the GMAX and RF shift toward a lower MPF from minute 1 to minute 30 of the walking bout with exoskeleton assistance compared to without (Fig. 1A). This seems to indicate an offloading of demand from the ankle to the hip with exoskeleton assistance.
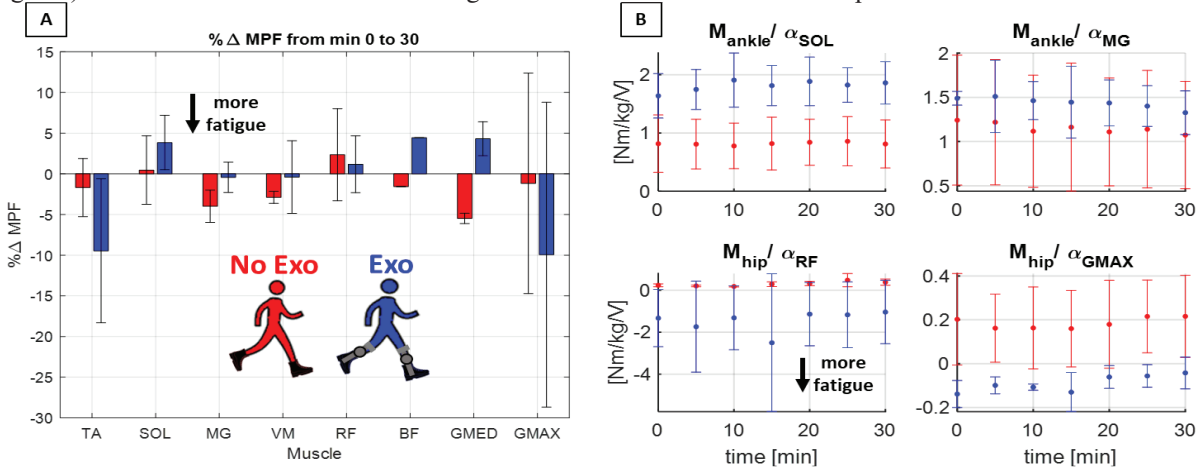

Figure 1: A) Mean Power Frequency (MPF) for 8 proximal and distal leg muscles. B) Joint moment per unit muscle activation at the ankle and hip.

Significance

Using the data set from this study, we will be able to better map the influence of exoskeletal assistance on muscle fatigue, providing a roadmap for the use of exoskeletons to mitigate fatigue not only to prolong walking capacity, but also to improve resilience to balance challenges and mitigate falls. We will also continue to investigate the effects of muscle level fatigue on locomotor stability during functionally relevant walking tasks, which is currently a gap in our exploration of fatigue and assistance in both younger and older adults, by incorporating various destabilizing perturbations during the 30-minute walking trial.

Acknowledgments

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ELASTIC EXOSKELETON INFLUENCE ON MUSCLE SPINDLE FIRING IN-VIVO

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Introduction

With significant advancement in wearable technologies for assisting locomotion and augmenting balance, there is a growing need to understand how such devices affect sensorimotor control of movement. Exoskeletons act mechanically in parallel to a joint (e.g., elastic exoboots) and may alter sensory feedback during in gait. Most models for the firing of muscle spindles, the sensory organs in muscles that signal joint motion, rely on joint kinematics, with the sensory firing rates driven by Muscle Tendon Unit (MTU) length and velocity. Due to the connective tissues in parallel and series to the intra/extrafusal muscle fibers, however, whole MTU kinematics likely do not well represent the force or length changes experienced by spindles. Recent work suggests that the contractile force and yank acting on the intrafusal fibers may more accurately predict spindle instantaneous firing rate (IFR) during passive MTU stretches [1]. Whether a muscle spindle responds to kinematic versus kinetic signals is critical to predicting how exoskeletons in parallel with the MTU would alter sensory signals. Based on a simple modeling framework we developed [2], if spindle firing depends on fascicle kinematics, we should expect firing to increase because of added spring stiffness in parallel during eccentric contractions. If it is based on fascicle kinetics, however, then firing should decrease. As we cannot measure spindle firing in humans, we translated our model to an acute rat preparation, where we are able to directly measure spindle firing and fascicle kinematics and dynamics in-vivo.

Methods

We directly measured how muscle spindle IFR is influenced by added exoskeletal assistance during an eccentric contraction in an acute rat preparation. We measured IFR through the dorsal root while imparting a predetermined sinusoidal length change (f = 2 Hz, amp = 1.5 mm) using a motor attached to the gastrocnemius MTU. We attached 4 springs (0.066 – 0.33 N/mm) in parallel to the medial gastrocnemius MTU and vary activation through the ventral root during stretch. Starting with a high parallel stiffness, passive stretch condition, we gradually decreased the parallel stiffness while increasing muscle activation so that the peak MTU+spring system force is matched across spring stiffnesses ($\pm$ 2.5%). The result was a length clamped, force-matched eccentric contraction across various exoskeleton stiffness conditions, akin to a locomotion cycle.

Results and Discussion

Figure 1A shows a select two stiffnesses and their effects on the contractile muscle force and fascicle length change, with the yellow bars indicating location of muscle stimulation. The IFR more closely aligns the muscle force over the stretch period than fascicle length. Figure 1B shows that the fascicle length (measured with sonos imbedded in muscle tissue) decreases with reduced parallel stiffness while the muscle force (measured by subtracting the spring force from the total MTU+spring force) increases. Figure 1C overlays the in-vivo measured IFR over the kinematic-based and kinetic-based model predictions, indicating a clearer correlation of the measured with the kinetic-based, rather than kinematic, IFR prediction.
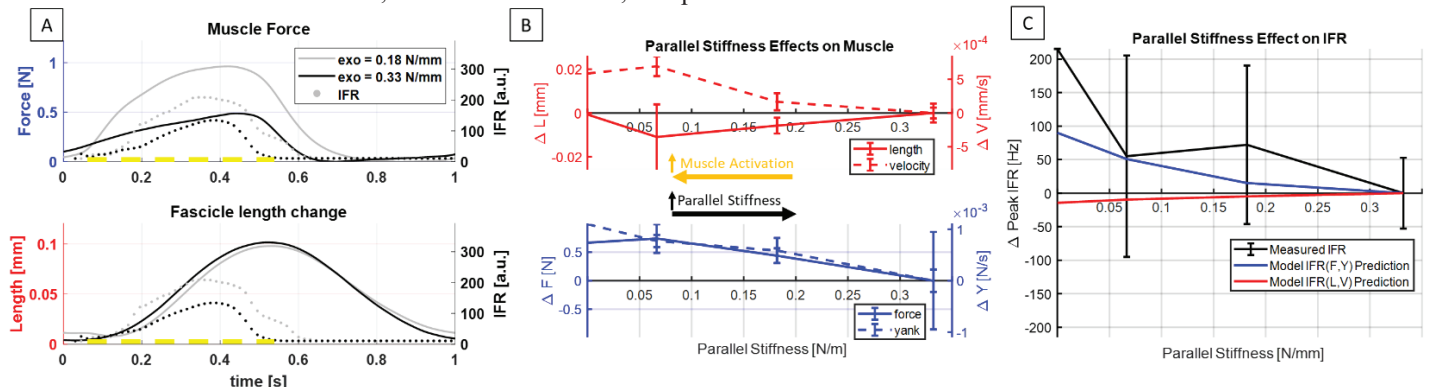

Figure 1: A) Muscle dynamics and IFR over 1 sinusoidal stretch across two exoskeleton conditions, B) Effects of varying exoskeleton stiffness across 4 conditions on kinematic (length and velocity change) and kinetic (force and yank change) factors experienced by the muscle fascicles, and C) Change in IFR measured versus model predictions. Results illustrated are from n=1 rat models.

Significance

As assistive devices become increasingly complex, an understanding of how they affect sensory feedback is critical if we are to develop exoskeletons more adept at addressing clinical challenges in motor learning and rehabilitation. Our approach allows for direct measurements of spindle response to known external loading, which can help reveal which muscle states contribute to spindle firing. Further collection in animal models, and eventually in humans, augmented with exoskeletal assistance, will help further illuminate the relationship between exoskeleton assistance, muscle mechanics, and resulting neural feedback [3].

Acknowledgments

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OFFSETTING THE LOAD: CAN EXOSKELETONS MITIGATE INJURY RISK DURING INDUSTRIAL LIFTING TASKS?

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Introduction: Exoskeletons have been shown to reduce user joint loading [1] and metabolic cost [2] during manual labor tasks. These findings have influenced the slow adoption of such devices in workplaces where manual labor tasks that involve twisting and lifting high loads are associated with high worker injury rates [3]. Both active and passive commercial exoskeletons are designed to offset joint loading by providing assistive torque either in parallel or perpendicular to the body. Such assistance reduces user muscle forces and activations [4]. Previous research has proposed that joint kinematics, joint kinetics, and muscle activity can indicate injury predisposition; however, few studies have investigated how exoskeletons affect joint and muscle loading in manual labor tasks. Further, the impact of exoskeletons on internal joint contact forces is largely unknown due to the inability to measure *in vivo* and the complex computations required for estimation. Our experimental protocol addressed this gap by recreating industry-relevant lifting conditions in-lab to understand joint and muscle-level demands. We hypothesized that an active knee exoskeleton and a passive back exoskeleton will reduce user back and knee extensor muscle activity while performing assisted lifts versus unassisted.

Methods: Ten participants lifted a 11.3 kg. weight during a symmetrical (0°- no turn) and asymmetrical (90° - rotational turn) task which varied in starting and ending lift height (Fig 1A). The weight was lifted from knee-to-waist (KW) height (ascension) and waist-to-knee (WK) height (descension). Participants performed each task 10 times. Participants wore an active knee exoskeleton, a passive back exoskeleton (HeroWear), and a no-exoskeleton case. We collected ground reaction forces, a full-body marker set, 16 inertial measurement units (IMUs) for segment orientation, and surface electrodes to record muscle activity via electromyography (EMG). We analyzed our data using OpenSim 4.0 and custom MATLAB scripts.

Results & Discussion: Contrary to our hypothesis, we found that assistance from the passive back exoskeleton, HeroWear, reduced back flexor (rather than extensor) peak muscle activity (rectus abdominis) by ~5% during symmetric lifting (Fig 1B). Compared to unassisted lifting, HeroWear's passive assistance reduced peak net back flexion moments (~10% symmetric and asymmetric), lateral bending (~20% symmetric), and axial rotation (~40% symmetric, ~10% asymmetric) (Fig 1C). In support of our hypothesis, we found that the active knee device reduced the peak muscle activity in the knee extensor (rectus femoris) [5] by ~20% and ~10% in symmetrical and asymmetrical lifting, respectively (Fig 1B). The peak net knee flexor moment only showed a ~5% decrease in asymmetric lifting (Fig 1C). Changes in the net joint moments imply that participants used different overall kinematic strategies to perform the given tasks. The elastic band design of the HeroWear and rigid interface of the knee exo may have constrained users' movement to operate in the sagittal plane, thus altering their lifting strategies. In lifting with fixed foot placements, asymmetric lifting could induce rotation and shear about the joints. Constrained motion with devices may add out of plane stability to help mitigate risks in asymmetric motions.

Significance: Our research suggests that both passive back and active knee exoskeletons may be able to mitigate injuries during lifting tasks. However, internal loading within joints (i.e., contact forces) cannot be captured from these data alone. Ultimately, we intend to use information about the external joint loads, muscle activity, and kinematic patterns to compute knee and back joint contact forces and study whether they are influenced by exoskeleton design. Until then, the observed reduced muscle activity with exoskeletons indicates that muscle forces may also be reduced [6], potentially lowering internal joint loads during lifting tasks and further reducing injury risk.
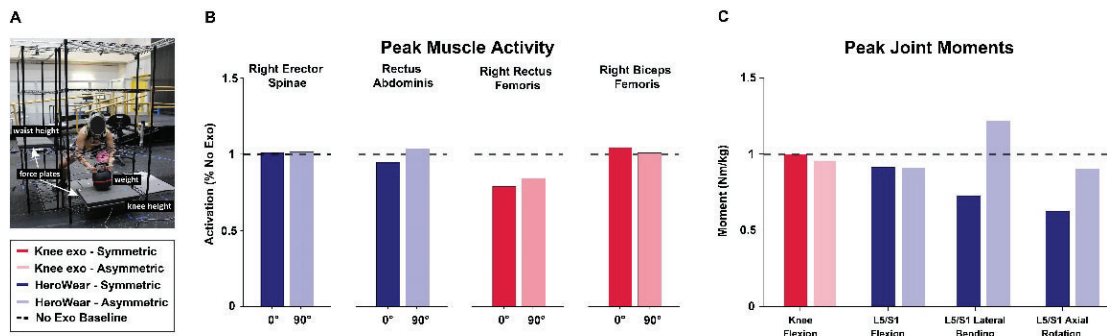

Figure 1: (A) Participant performing a lifting task. **(B)** Peak muscle activity normalized to the peak no exo muscle activity in symmetric (0°) and asymmetric (90°) lifting. **(C)** Peak net joint moments (exoskeleton + biological moments) in symmetric (0°) and asymmetric (90°) lifting. All values are normalized to the no-exo peak net joint moment. Each bar represents averaged ascension and descension conditions.

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WEARING HIGH HEELS REMODELS LEG MUSCLE-TENDONS AND IMPROVES WALKING ECONOMY

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Introduction: Each morning people wake up and put on clothing and footwear, and a assistive devices (*e.g.,* prostheses). Most of these items improve user physical function, convention misses an opportunity to use wearables to alter muscle-tendon structure and after the item is removed from the body.

Cross-sectional- and modeling-studies suggest that the habitual use of high-heeled shortens calf muscles by $\geq$9-14% and stiffens Achilles tendons by 22% [1]. Shorter leg theoretically enable people to walk using less metabolic energy (walk more because shorter muscles have the same force-producing capabilities as longer muscles housing fewer energy consuming sarcomeres in-series ($\downarrow$length), and stiffer tendons shortening per unit force production (x=F/$\uparrow$k) [2]. Considering that calf muscles produce most of stance while expending ~27-40% of the body's metabolic energy during walking and stiffening the Achilles tendon may reduce a person's metabolic energy expenditure Accordingly, we hypothesized that the habitual use of high heels would reduce a person's expenditure during walking in flat-soled footwear.

Methods: Eight young adults agreed to wear custom high heels as their everyday shoes heels placed participant ankles at 104° during standing. Throughout the protocol, log their daily steps with and without their high heels. Immediately before and after 12-heels, we conducted laboratory testing that involved treadmill walking at 1.3 m/s in flat-well as multiple ankle plantarflexion trials on a dynamometer at different joint angles.

Results: Participants varied in the amount that they wore their high heels (daily steps in ± 1257 steps/day; range: ~0 to 3704 steps/day). Participants who took more daily steps in heels stiffened their Achilles tendons and improved their walking economy in flats; supporting our hypothesis. For every 1000 steps/day that participants averaged wearing high heels, their Achilles tendon stiffness increased 7-8% (β= 19 kN/m; p=0.008) and their net metabolic power during treadmill walking decreased 3% (β=-0.09 W/kg; p=0.004) (Fig. 1). ΔTendon stiffness did not correlate with Δnet metabolic power during walking (r=-0.64; p=0.086) (Fig. 2). Moreover, post-hoc analyses suggest that averaging >1k steps/day in high heels is necessary to reduce net metabolic power during walking (p=0.018). Participants who averaged >1k steps/day in high heels reduced their net metabolic power during walking in flats by 9±3% (avg ± sd) (n=4; red symbols), whereas participants who averaged <1k steps/day in high heels consumed 5±7% (avg ± sd) more metabolic power during walking in flats post intervention (n=4; grey symbols) (Fig. 1).
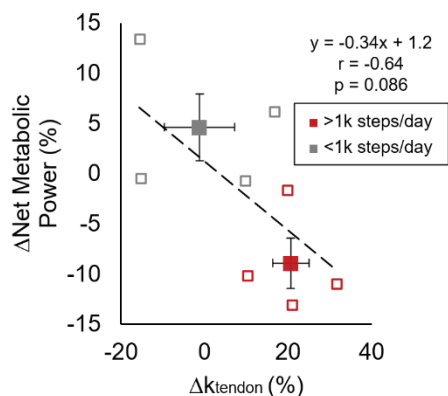

Figure 1. (a) Δnet metabolic power during walking at 1.3 m/s and (b) ΔAchilles tendon stiffness versus average steps in high heels per day over 12-16 weeks. Dashed lines indicate linear regressions on individual data (open symbols). Closed symbols represent Avg. ±SE values for participants who took >1k and <1k steps/day in high heels, respectively.
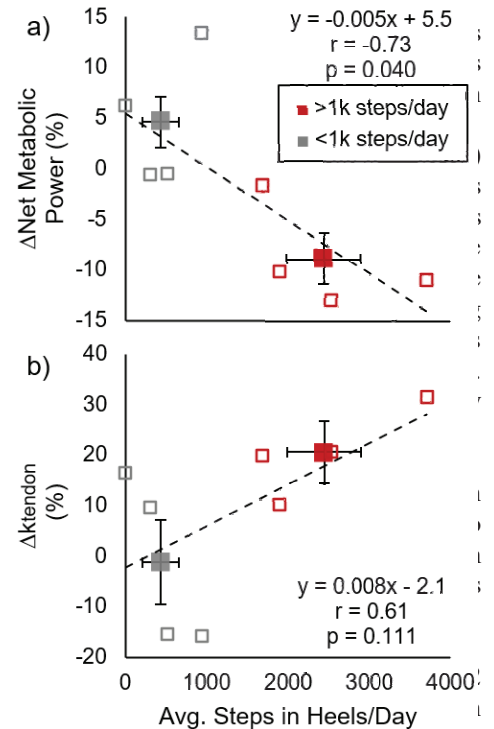

Figure 2. Δnet metabolic power during walking at 1.3 m/s versus ΔAchilles tendon stiffness. Dashed line indicates linear regressions on individual data (open symbols). Closed symbols represent Avg. ±SE values for participants who took >1k and <1k steps/day in high heels, respectively.

Discussion: Participants who took >1000 steps/day in high heels for 12-16 weeks reduced their metabolic energy expenditure during walking in flats. While we are still analysing the data to uncover which physiological changes contributed to the more economical walking following habitual high heel use, increased Achilles tendon stiffness may have contributed. Despite not reaching statistical significance with 8 participants (p=0.086) (Fig. 2), the moderate correlation between ΔAchilles tendon stiffness and Δnet metabolic power (r=-0.64) supports with the notion that increasing tendon stiffness enables in-series muscles to produce force more economically [2]. Further, habitual high heel use may have also shortened calf muscle fascicles [1] and/or elicited neuromechanical changes that contributed to the improved walking economy following high heel use.

Significance: The strategic design of everyday wearable items can improve user morphology and function *after* items are removed from the body. Because muscle-tendons take weeks to remodel, modifying leg muscle-tendon structure using high heels may improve user walking economy for multiple weeks after the high heels are last worn. Our study aims to inform the design of wearable items that can improve user physical function and mobility while the item is worn as well as after the device is removed from the body.

References: [1] Csapo et al. *J Exp Biol* 2010, [2] Krupenevich et al. *Gerontol* 2022, [3] Umberger & Rubenson *Ex Sport Sci Rev* 2011.

HOP, SKIP, AND A JUMP: INVESTIGATING WHY PEOPLE JUMP IN RESPONSE TO WALKING PERTURBATIONS

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Introduction: What causes people to fall in the real world? Despite a vast amount of literature investigating balance metrics in response to perturbations, it may be the strategy that is required for balance recovery that causes people to fall rather than the magnitude of imbalance itself. In our recent work, we studied the balance and recovery strategies used to respond to perturbations that varied in magnitude, direction, and timing [1]. As a result of studying a thorough sweep of these variables (96 conditions), we observed a niche set of conditions that elicited an extreme balance recovery strategy; in over 20% of responses to some conditions, individuals jumped. Here, we took a closer look at the jump responses in our data set by analyzing two situations that we hypothesized would elicit a jump response and the three jump mechanisms that we observed.

Methods: We used our previously published data set; 11 participants walked while being exposed to ground perturbations that varied in magnitude, direction, and timing [1]. We hypothesized that **H1)** participants jumped to avoid a **collision** of the swing limb with the stance limb, typically elicited during a crossover step (Fig. 1A). We quantified this using the velocity vectors of the swing foot in the 150 ms leading to the jump; we classified a projected collision if the velocity vector of any swing foot marker was projected to collide with the region defined by the stance foot markers. If this was not the case, we hypothesized that **H2)** remaining jumps occurred if the required step width was **too narrow** and fell outside the capabilities of the participant (Fig. 1B). To quantify this, we fit a participant-specific center of mass-driven model [2] to the four steps after each non-jump perturbation trial. We used this model and center of mass mechanics leading up to the jump to project the required step width had the participant not jumped. We quantified three mechanisms that individuals use to jump (Fig. 1C); 1) a lateral **skip** strategy involves pushing off of the stance foot and landing on that same foot lateral to the original position, 2) a **foot replacement** strategy involves hopping into the air with the stance foot and landing in the same location with your swing foot, and 3) a **leap** strategy involves hopping into the air with the stance foot and landing anteriorly with the swing foot.
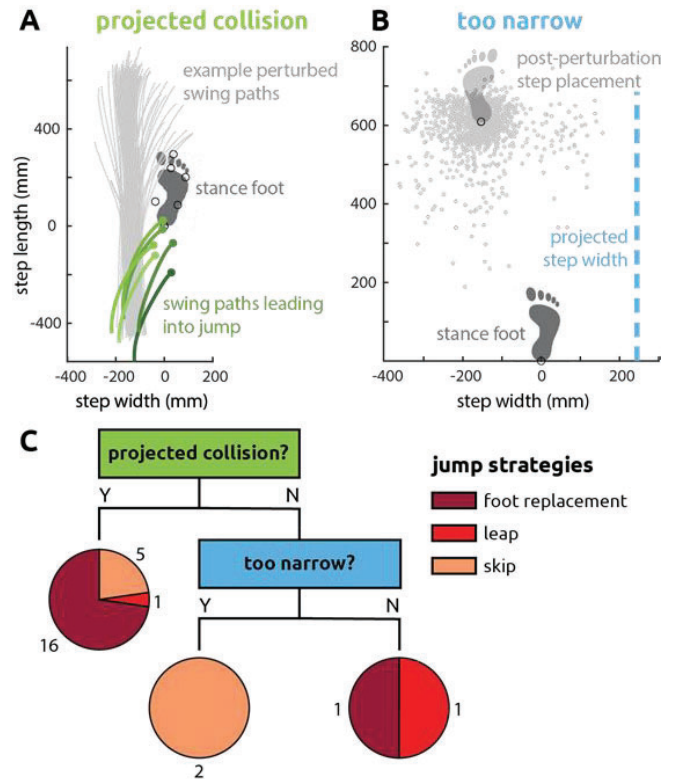

Figure 1: (A) Lines show swing foot heel marker relative to stance foot; grey lines are non-jump responses, green lines are swing phase leading up to jump. **(B)** Grey dots show non-jump post-perturbation steps, blue line shows a trial's projected step width from the linear model, which is a narrower step than any successfully executed step by the participant. **(C)** Decision tree showing the jump strategies used in response to projected collisions and too narrow steps.

Results & Discussion: Of the 26 trials with jump responses analyzed from the data set, 22 trials were projected to have a limb collision. In the jumps that followed the projected limb collision, 16/22 used a foot replacement strategy, 5/22 used a skip strategy, and 1/22 used a leap strategy. In the remaining 4 trials that did not include a projected limb collision, 2/4 were projected to require too narrow of a step. In both of these trials, individuals used a skip strategy. In the remaining 2 trials that did not present a collision or too narrow of a step, the foot replacement and leap strategies were both used once. Broadly, this work identifies potential limb collisions during a narrowing step maneuver as the leading cause of jump responses following perturbations. In these situations, participants dominantly reacted by using a foot replacement strategy; in addition to preventing a collision, this strategy effectively turns a narrowing step into a widening step, which may set up the participant to a wider range of maneuvers on the subsequent step. The lateral skip strategy effectively does the same, which is executed for the remaining steps that are too narrow.

Significance: Here, we identified some of the more demanding balance recovery mechanisms that have been reported, with only two other studies that we are aware of reporting jump responses [3,4]. The perturbation conditions that caused jumps could be a useful tool to study highly destabilizing scenarios, especially those that may cause a fall in balance-impaired individuals. These responses are also important to consider in the development of wearable robots, as these strategies may present edge cases for existing control architectures. Lastly, these responses pose an interesting stance/swing limb constraint challenge that should also be considered in bipedal robotics.

References: [1] Leestma et al. (2023), *J Exp Biol* [2] Joshi, Srinivasan (2019), *J R Soc Interface* [3] Eveld et al. (2020), *Mid-South Biomechanics Conference* [4] Pijnappels et al. (2004), *J Biomech*

DOES THE RELATIONSHIP BETWEEN WHOLE-BODY ANGULAR MOMENTUM AND STEP PLACEMENT CHANGE IN INDIVIDUALS POST-STROKE?

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Introduction: Individuals post-stroke typically have impaired balance and are at an increased risk of falling [1]. Gaining a more thorough understanding of how balance recovery strategies are altered post-stroke could drive more effective rehabilitation strategies and improve the function of assistive devices for a post-stroke population. Here, we aimed to investigate how step width is modulated in response to frontal plane instability in neurotypical and post-stroke individuals. We hypothesized that H1) neurotypical individuals would display a weaker correlation between balance and step placement compared to post-stroke individuals, as post-stroke individuals typically have weaker ankles, which inhibit the ankle strategy that neurotypical individuals use to combat small perturbations. We also hypothesized that H2) correlations would be stronger for steps where the non-paretic limb is in stance in comparison to the paretic limb, due to the higher joint moment demand on the stance limb in comparison to the swing limb when making step width adjustments.

Methods: One post-stroke participant walked on a treadmill at 0.8 m/s while being exposed to ground translation perturbations. We applied 5 cm perturbations that varied in direction (anteroposterior, mediolateral) and onset timing (double stance; early, mid, late single stance). We applied each perturbation condition 3 times to the paretic and non-paretic limbs. We collected a full-body marker set and identified gait events using a kinematic method [2]. We calculated integrated frontal whole-body angular momentum (WBAM) using OpenSim and custom Matlab scripts. We calculated step width using the mediolateral distance between heel markers. We also used our previously collected open-source data set to provide 11 neurotypical participants with condition-matched perturbation trials for comparison [3]. We evaluated the correlation between integrated frontal WBAM and step width in the perturbed and recovery steps.

Results & Discussion: We expected low R^2 values for neurotypical individuals, as lateral ankle strategy is typically sufficient to combat minor amounts of instability whereas step width modulation is required for more severe perturbations. However, we still saw moderately strong correlations in the perturbed ($R^2=0.46$) and recovery ($R^2=0.20$) steps, which were roughly equal to or greater than the paretic and non-paretic comparison; this did not support H1. In the perturbed step, there is a stronger correlation in non-paretic stance steps ($R^2=0.48$) than in the paretic stance steps ($R^2=0.27$), supporting H2. However, in the recovery step, there was a stronger correlation in paretic stance steps ($R^2=0.05$) than in non-paretic stance steps ($R^2=0$), which does not support H2. These results are preliminary and only include a single stroke participant; we have collected and are analyzing additional post-stroke participants to incorporate in this analysis.

Significance: Understanding how individuals' balance and recovery strategies are affected following a stroke could enable therapy strategies that target more comparable balance responses to a neurotypical population. Additionally, uncovering asymmetries and insufficiencies in post-stroke balance recovery could help inform controllers for wearable robots that alter assistance between the paretic and non-paretic limbs or customize assistance to individual post-stroke users.

References: [1] Weerdesteyn et al. (2008), *JRRD* [2] Zeni et al. (2008), *Gait Posture* [3] Leestma et al. (2023), *J Exp Biol*
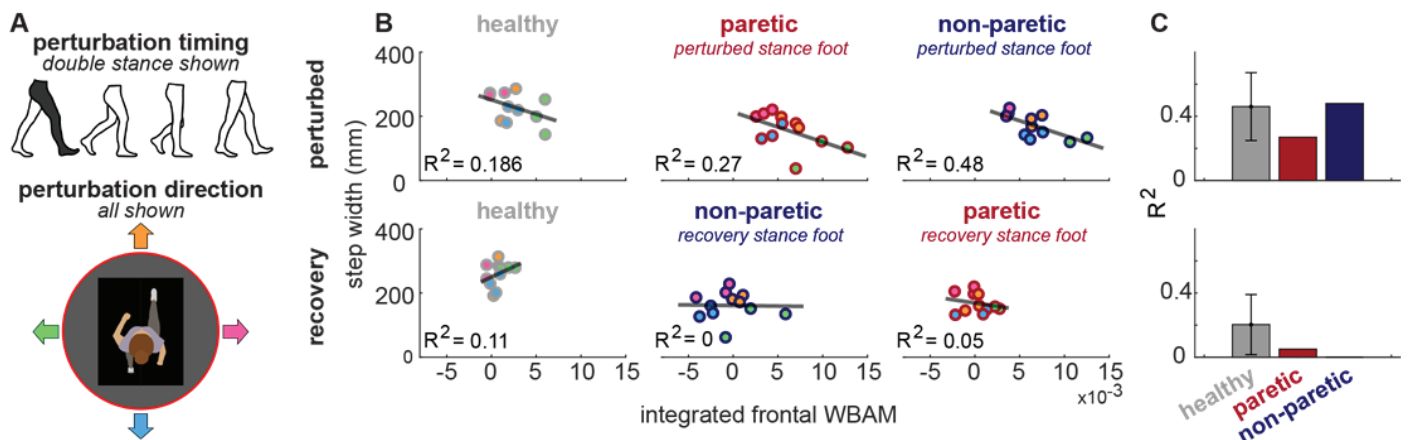
Figure 1: (A) Data shown in plots are for the double stance onset time across all perturbation directions. **(B)** Correlations between integrated frontal whole-body angular momentum (WBAM) and step width during the perturbed (top row) and recovery (bottom row) steps; grey outlines are for a representative healthy participant, red outlines indicate the paretic foot is in stance, navy outlines indicate the non-paretic foot is in stance. **(C)** The R^2 value for the relationships in the plots. The healthy group (grey) contains the mean and standard deviation for R^2 values from 11 participants, the paretic (red) and non-paretic (navy) data shows the R^2 value for a single stroke participant.