COMPARING GAIT QUALITY METRICS ON THEIR ABILITY TO DISTINGUISH BETWEEN A TUNED AND UNTUNED ROBOTIC PROSTHESIS

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Introduction

What does it mean to have good gait? Many existing measures of overall gait quality are based on different philosophies and applications with no consensus "gold standard." This can make it difficult for clinicians that are often restricted by time and subjectivity when treating various gait pathologies. In the case of powered robotic prostheses, objective, discrete parameter changes are necessary in the tuning process to achieve an improved gait pattern. Having clear gait quality metrics that inform such tuning changes can be a powerful tool to aid clinicians in making effective changes to a patient's gait. A well-tuned powered robotic prosthesis has the potential to improve metabolic cost, joint loading, and overall mobility for those with lower limb amputations[1]. Our aim was to characterize the ability of different gait quality metrics in distinguishing the effects of tuning a powered prosthesis.

Methods

Eight subjects with below knee amputation along with nine healthy control subjects provided informed consent according to the Georgia Institute of Technology IRB prior to study enrollment. Mean age, height, weight, and gender did not substantially differ between the two groups. A certified prosthetist fitted and aligned subjects with an amputation to a powered below knee prosthesis emulator (Humotech PRO-001)[2]. They then walked on an instrumented split-belt treadmill at 1.0 m/s as the clinician tuned the prosthesis using observational gait analysis and subject feedback. We recorded 3D kinematics and kinetics (Vicon, Visual3D) following each tuning parameter change for 15 s until the patient reported feeling comfortable and the gait was clinically assessed to be visually appropriate and considered "tuned". Preliminary data analyzed from six of the subjects with amputation are presented in this abstract. Gait quality was assessed using five known gait indices: impulse asymmetry[3], lateral sway[4], Prosthetic Observational Gait Score (POGS)[5], Gait Deviation Index (GDI)[6], and Gait Quality Index (GQI)[7]. Mean control group data were used to calculate the GDI and GQI for subjects using the prosthesis. The untuned and tuned conditions on the preliminary data set were compared using effect sizes estimated by Cohen's *d* with a 95% confidence interval.

Results and Discussion

The tuning process had a medium sized effect on impulse asymmetry, POGS, and GQI while showing a small to negligible effect on lateral sway and GDI (Table 1, Fig. 1).

The observational gait assessment integral to the manual tuning process may have led to the effect indicated when using POGS, which is a time-consuming visually scored rating scale. Impulse asymmetry and GQI are gait quality metrics that assess ground reaction forces and joint power data, respectively. The effects observed when using the impulse asymmetry and GQI metrics likely reflected improved kinetics due to changes in push-off power provided by the prosthesis emulator.

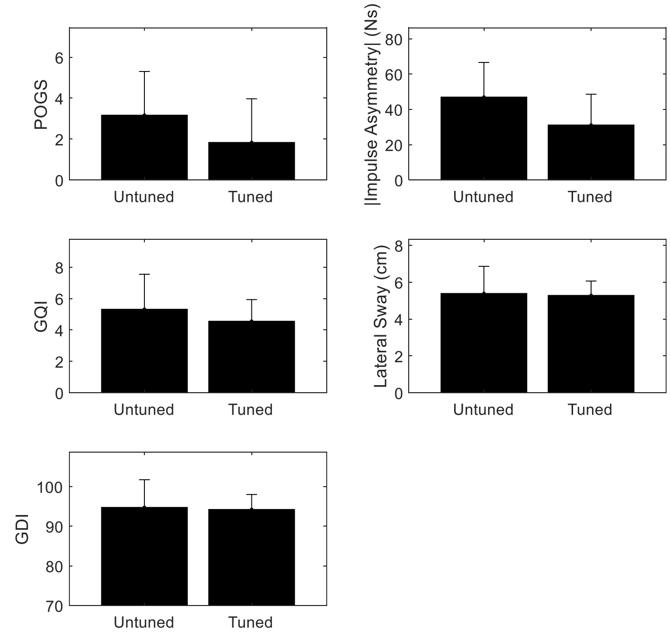

Figure 1: Untuned and tuned mean ± s.d. values for each gait metric.

Our results indicate that an observational gait metric along with metrics detecting the effects of changes in prosthesis force production were best at identifying gait improvements due to manually tuning a powered below knee robotic prosthesis.

Significance

Tuning powered prostheses is a subjective, time-consuming process. While observational gait assessments are most widely used, metrics that leverage gait kinetics can also be valuable and could potentially lead to more rapid, automated tuning methods.

Acknowledgments

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Table 1: Cohen's *d* values and 95% confidence intervals for each gait metric.

	POGS	Impulse Asymmetry	GQI	Lateral Sway	GDI
***d* [95% CI]**	0.678 [-0.486, 1.842]	0.512 [-0.638, 1.661]	0.505 [-0.645, 1.655]	0.110 [-1.022, 1.242]	0.103 [-1.029, 1.235]

A PASSIVE HIP FLEXION DEVICE MAY IMPROVE STABILITY DURING PERTURBED WALKING

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Introduction

A major open question in exoskeleton research is how to design devices that improve stability during walking. Current approaches targeted to users with high levels of volitional control focus on providing motor-driven assistance at the hip joint (*e.g.*, [1]). However, passive devices may provide a lightweight, low-profile source of stabilizing assistance for perturbations that induce a forward pitch followed by a hip flexion response, such as forward slips or trips.

Here we present preliminary work investigating the influence of a commercial elastic hip flexion device (Moveo Exoband) on stability following rapid, transient unilateral treadmill belt accelerations designed to induce forward pitch during walking. [2] We quantified stability as the range of whole-body angular momentum (WBAM) over a stride, which is considered tightly regulated during unperturbed walking. [3] We hypothesized that increasing Exoband stiffness would improve stability during perturbations as evidenced by a decreased range of WBAM.

Methods

1 subject (M, 30 y/o, 75.8 kg, 188.5 cm tall) walked on a split-belt treadmill at 1.25 m/s in an instrumented version of the Exoband and experienced 20 transient unilateral belt accelerations (early and late stance perturbation timings, left and right legs, 5 repetitions; [2]) in each of 4 stiffness conditions. Exoband stiffness was altered using different compression springs (No spring, Low=4.9 N/mm, Medium=7.2 N/mm, High=10.9 N/mm). Exoband flexion torque (Fig B,C) was calculated using load cells in-series with the springs. The lever arm of the springs relative to the hip joint was calculated from motion capture. WBAM was calculated in OpenSim 4.0 using a full body musculoskeletal model. [4] Data were averaged across legs and repetitions and are presented for the stride before (S-1), during (S0), and after (S+1) the perturbation.

Results and Discussion

Across all stiffnesses, a forward pitch was induced by the perturbation in either early or late stance, as demonstrated by a shift to negative WBAM values on the perturbed stride (S0; Fig 1 D,E). The Exoband stiffnesses that maximized stability was different between the two perturbation timings (Fig 1 F,G) – for early stance perturbations, in agreement with our hypothesis, the high stiffness Exoband condition resulted in the smallest mean WBAM of all conditions due to a reduced backwards pitch in recovering from the perturbation. However, in contrast to our hypothesis, for late stance perturbations, the medium Exoband stiffness resulted in the smallest mean WBAM due to decreased forward pitch during the perturbation. Further, the effect of stiffness on WBAM was more muted for the late vs. early stance perturbation timings.

Overall, these preliminary findings indicate the timing of perturbations is critical for determining the mechanism and extent to which passive hip flexion assistance improves stability.

Significance

This is the first study demonstrating the potential of passive assistance to improve locomotor stability. Work is currently underway to collect a larger (N=11) data set and understand the muscle-level implications of such assistance in unstable contexts.

Acknowledgments

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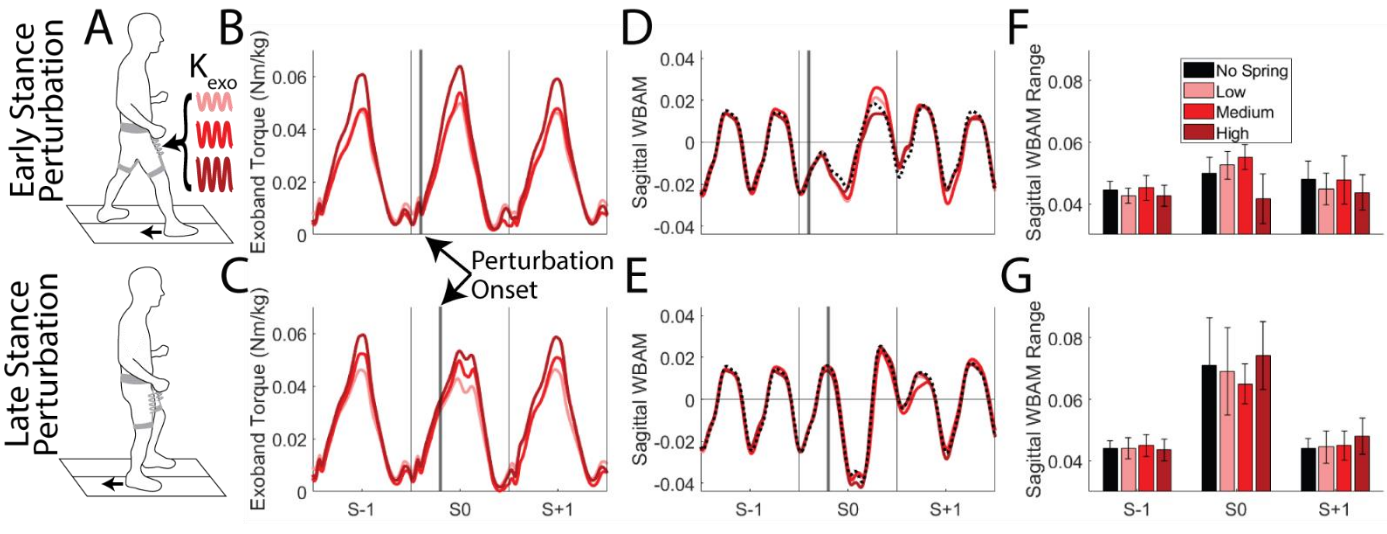

Figure 1: Top/Bottom row: Early/Late stance perturbations, respectively, for strides before (S-1), during (S0), and after (S+1) the perturbation. **A –** Methods overview. **B,C** – Flexion torque provided by Exoband. **D,E** – Sagittal whole body angular momentum normalized to height, mass, and walking speed. **F,G** – Range of normalized sagittal WBAM

METHOD FOR PREDICTING 3D GROUND REACTION FORCES UNDER VARIOUS LOADING PATTERNS

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Introduction

We are in the process of building a powered posterior walker to improve the utility of the device by relieving the cost of pulling the walker, while maintaining the stability benefits of using the walker. Additionally, prior work has shown that applying a horizontal forward assist force of 8% body weight can reduce the metabolic cost of walking on a treadmill by 34.7%.[1] We aim to transfer this assistance to over-ground walking by controlling the posterior walker to apply this assistance to the user.

To determine the most beneficial assistance method for our walker to apply, we are developing a simulation framework that can predict ground reaction forces (GRF) and accommodate additional external forces, while minimizing the kinematic constraint forces. In this case, kinematic constraints must be just stiff enough to maintain stability, while flexible enough to predict GRF and changes in GRF as different external forces are applied. Predicting changes in GRF will allow us to calculate the joint power of the user under different assistance conditions. Ultimately, we can use this method to determine when and how the walker should apply assistance to the user.

<u>The work presented here develops a method of predicting GRF of human walking in 3D, such that GRF are free to change when additional external forces are applied to the simulation.</u>

Methods

A full-body musculoskeletal model with 29 segments and 36 DoF was developed in OpenSim, based on the Hamner 2010 model, and scaled to the subject being modelled. Ground contact was modelled by 3 spherical Hunt-Crossley elements on each foot, with stiffness of 3.06 MPa and dissipation coefficient of 4. We implemented this model into the OpenSim Moco framework [2] in a torque driven simulation where the cost function is set to minimize: (1) the deviation from the reference states being tracked and (2) the control effort, with a penalty on controller effort at pelvis to ground controllers by weighting these controllers 10x all other controllers in the system. No information about the GRF recorded in the trial the motion came from was provided to the simulation. This framework was used to predict the GRF in three over-ground walking trials. The resulting GRF were filtered at 10 Hz. RMS error was calculated between the Moco predicted GRF and GRF recorded on force plates (Bertec) during the respective walking trial.

Results and Discussion

This framework allows us to predict 3D GRF, as shown by agreement with the recorded GRF (***Table 1, Figure1***).

Table 1. *RMSE of predicted force vs. recorded force.*

RMSE (N)	**Vertical**	**Anterior**	**Medial**
Avg. Error ± St.Dev.	92.3 ± 36.7	37.1 ± 9.4	12.0 ± 3.3

Further, using this method results in kinematic constraint forces that are much smaller than those utilized in inverse dynamics methods, with the mean vertical kinematic constraint force using this framework averaging 1% of the mean vertical kinematic constraint needed in an inverse dynamics method (***Table 2***). This will give us the flexibility needed to see changes in GRF when we implement assistance into these simulations.

Table 2. *Translational external kinematic constraint forces.*

Force (N)	**Vertical**	**Anterior**	**Medial**
Inv. Dynamics	160.7 ± 115.7	9.2 ± 2.8	3.9 ± 3.2
Moco	1.8 ± 1.1	3.3 ± 0.7	0.2 ± 0.2

With this framework in place, we will be able to simulate countless possible control methods for assistance from the powered walker by varying: duration and amount of assistance, timing of assistance in the gait cycle, and location on the user assistance is applied. This method will provide us the GRF needed to calculate the joint kinetics through the body, and ultimately the efficiency of walking with that assistance method.

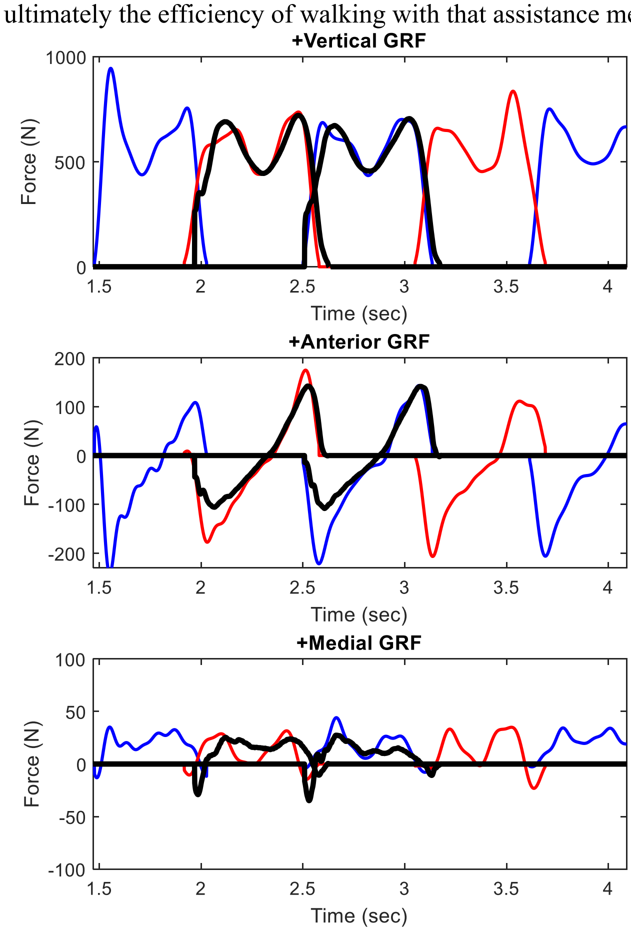

Figure 1. Ground reaction forces for representative trial, with Moco results in **blue (left)** and **red (right)** against the recorded GRF in black.

Significance

The ability to predict GRF allows for the investigation of optimal assistance methods. We will be able to try vastly more methods of how a powered walker could assist its user through simulation, than one could ever hope to test in a clinical setting. This will allow us to explore many different combinations and timings of assistance and narrow down this space, so as to only test the most promising assistance methods in the lab.

Acknowledgments

Thank you to the UVA Motion Analysis & Motor Performance Lab members.

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DESIGN AND EVALUATION OF A BIMODAL PEDIATRIC PROSTHETIC FOOT FOR WALKING AND RUNNING

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Introduction

Children frequently alternate between walking and running as they switch between activities during play. However, these transitions are a problem for children with lower limb prostheses. While many have access to a daily-use prosthesis (DUP) and a specialized running-specific prosthesis (RSP), it is impractical for a child to switch devices as frequently as they switch activities. This leads to an increase in non-task-specific device use (e.g., walking in an RSP or running in a DUP), which is associated with performance detriments and poorer user satisfaction [1]. These limitations can deter children from participating in group play, which may hinder their physical, social, and psychological development [2].

While many prostheses can be used for both walking and running [3], the device dynamics are not optimized for both tasks. One bimodal prosthesis on the market, the Ottobock Challenger, is not self-contained (the removable heel must be stored separately when running), limiting its practicality for children. In response, we propose an easily adjustable, bimodal prosthetic foot, i.e., a prosthesis optimized to the task demands of walking in one mode, and running in another, that requires minimal time and force application to switch modes. In this abstract, we outline several design requirements and present a prototype.

Methods

In order to test if this device can perform as both an optimal DUP and an RSP, we established design criteria focused on important characteristics from each type of device.

As a DUP, the device needs to have similar walking dynamics to other common DUPs and provide a large footprint. The device's walking mode should have the same compliance as a typical DUP (6-9 Nm/deg) [4] to ensure the device would be comfortable for hours of use throughout the day. The device should provide a broad footprint (the area of contact between the device and the ground) to support the shifting center of pressure while standing. The footprint should be comparable to a child's shoe sole area.

As an RSP, the device should have a high energy storage and return ratio, clearance at the heel, and be able to withstand the impact forces associated with running. To ensure that the device can mimic the high energy return of an RSP, the running mode should have low hysteresis and a capacity for storing energy comparable to that of a typical RSP, 0.19 J/(kg*m) [5]. To avoid the heel scuffing on the ground during the flight phase, the running mode should provide increased clearance at the heel. The force of impact while running can range from two to three times the runner's bodyweight; to ensure robustness, the running mode components should be suitable for users up to 90 kgs.

After establishing these criteria, and additional secondary criteria (not listed), we designed and built a computer model and physical prototype of a self-contained prosthesis that can switch activity modes.

Results and Discussion

There are two key features to our device that enable it to be optimal for both running and walking: the Parallel Spring and the Collapsible Heel (Figure 1). With a simple twist of the outer barrel, the Parallel Spring Strut will either engage or disengage an internal compression spring. When disengaged, the spring will act transparently and will not affect the stiffness of the device such that it behaves like a typical DUP. When engaged, the spring will increase the overall stiffness of the device and enhance its ability to store and return energy such that it mimics the energetic properties of a commercially available RSP. The internal spring is swappable and the device can hold springs capable of storing more than 40 J of energy (yielding over 0.2 J/(kg*m) for a 90 kg user with a 2 m stride length).

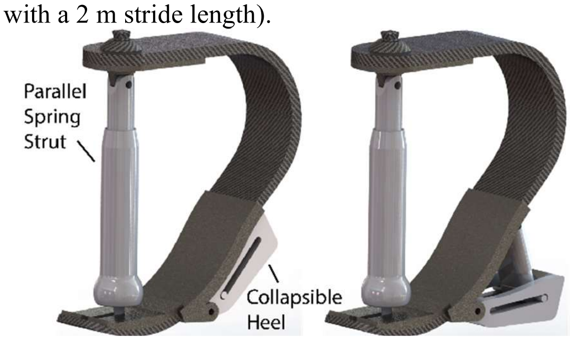

Figure 1: Rendering of the prosthetic prototype with the Parallel Spring Strut and the Collapsible Heel (collapsed on the left, deployed on the right).

When the Collapsible Heel is deployed, the footprint doubles from 78 cm^2 to 161 cm^2 (comparable to that of a men's US size 6 shoe), making the device comfortable for both standing and walking. Additionally, the springs in the Collapsible Heel's weight-bearing member mimic the stability and compliance provided by a leaf spring heel found on commercially available DUP. When collapsed, the heel rests against the posterior of the blade and provides increased clearance while running (6-7.5 cm, depending on orientation), mimicking an RSP.

Prior to the conference, we plan to collect preliminary data from lower limb prosthesis users aged 8-17 (N=3). After walking, running, and switching modes on the device, participants will complete a subjective survey where they will report their satisfaction with the device as a DUP, as an RSP, and the ease of switching between modes. Quantitative data regarding the compliance of the foot in walking mode and the energetic storage properties in running mode will also be taken at this time.

Significance

This prototype is designed to increase a child's satisfaction with their prosthetic limb as both a DUP and RSP, and could increase the likelihood that children engage in group sports or play.

Acknowledgments

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ADJUSTABLE EFFORT BIKE PEDAL SYSTEM FOR LEG REHABILITATION

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Introduction

Cycling, as an exercise therapy, is used in clinics for strengthening muscles, rehabilitating leg ailments, decreasing pain, and improving physical function of knee joints [1]. Conventional bicycles, both stationary and outdoor, still present challenges for post-surgery recovery or pain treatment of osteoarthritis [2]. There is a potential value in a device that can provide the patient with asymmetrical assistance while cycling with the goal of providing more power to the injured leg and not "over-assist" the healthy leg. Therefore, the goal of present project was to develop an instrument bicycle that provides assistive torque to the injured leg during pedaling. The developed bicycle tracks the position of the pedals and measures the torque and foot force real-time. It would be of interest to examine what effect such a device would have on muscles, bones, and cartilage functions in injury recovery, and osteoarthritis pain management.

Methods

A commercially available electric bike is outfitted with a crank angle sensing device, a pedal force measurement device, and a microcontroller that feeds a desired throttle signal to the bike motor. The electric bike was mounted to a stationary resistance trainer. The patient will either pedal the bike mounted on a stationary bike trainer or cycle in a controlled outdoor environment. The motor is activated when the crank angle is within a user-defined range, identified by the hall sensors, so that the motor biases the assistance to only one leg. The active range of the motor is adjustable depending on patient physiology or the advice of the patient's physical therapist. A pilot test was conducted on a healthy subject (male, age 35, height 180cm), where the participant pedaled at 40 rpm and 65 Watt power for 2 minutes when the motor was off (control), followed by a 2min pedaling at the same rpm and power with active control of motor (treatment). The EMG of vastus lateralis muscle on both legs and pedal force were recorded. In the active phase, the motor applied assistance torque when the left leg pushed down the pedal, i.e. 0 < left crank angle < 180. The T-test was conducted to compare the results of the control and treatment.

Figure 1: Subject operating the device during motion capture trial

Results and Discussion

Initial results from the pilot tests showed that when the device was set to (treatment) from the top of the pedal stroke to the bottom (0 < left crank angle < 180) at moderate motor power, there was a significant attenuation ($P<0.01$) in sEMG magnitude and pedal force compared to when the motor was off (control), while participant's kinematics remained unchanged (Figure 2).

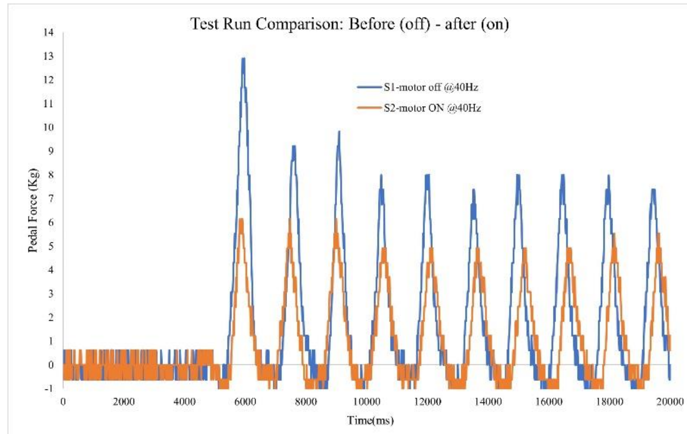

Figure 2: Pedaling force of left leg @ 40 rpm & 65W power. Motor OFF: (Control), Motor ON: (Treatment) from first 20 seconds of a pilot test.

Significance

The result of this study showed that the developed device can provide synchronized asymmetric power to assist a potentially impaired leg while facilitating data capture. Such work has the potential to enhance patient recovery experience and open possibilities for further research to use it for outdoor activities such as cross-country and trail ridings.

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Assistance within electromechanical delay between muscle activation and contraction, a case study

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Introduction

Demyelinating disease can lead to a loss of muscle coordination, reduced postural control, tremor, and slower reaction times [1]. To expand available treatment options for more rapidly fatiguing patients, intervention strategies utilize external electrical stimuli [1,2]. These electrical stimuli excite existing axons which is thought to trigger remyelination. They can improve either the functional capabilities of peripheral muscle, improve motor learning capabilities, or quicken the speed of improvement from rehabilitation [2,3]. However, the effectiveness of these treatments is somewhat inconsistent [1]. We suspect that the inconsistencies of the electrical stimulation may be due to an absence of congruent mechanical sensory input. Alternative intervention strategies have attempted to leverage mechanical stimulus to modulate myelin formation via whole-body vibration [4]. However, application of whole-body vibration has inconsistent effects and forgoes the potential mechanism of excitation seen via electrical stimulus. We present a preliminary intervention that facilitates an increase in the muscle activation in conjunction with a mechanical sensory stimulus. We hypothesized that perturbing elbow extension following initial muscle activation, but prior to the force producing contraction, would illicit higher overall subsequent muscle activation and quicker reaction times than elbow extension alone.

Methods

One healthy, young subject participated in this study. Once receiving a somatosensory cue underneath the left heel, the participant conducted 90 degrees of lateral, right elbow extension while resting their shoulder on a table. Surface EMG sensors (Noraxon USA, Inc. Scottsdale, AZ) was placed on the lateral triceps brachii and the lateral biceps brachii. EMG was recorded at 1000Hz. Upon initiating movement, identified via surface electromyography, four different forces were applied to the participant's arm as different study conditions: an assistive force ~20ms prior to and ~20ms following the participant's muscle contraction, and a perturbation (pull opposing elbow extension) ~20ms prior to and ~20ms following the participant's muscle contraction. Each condition set (assistance, slow assistance, perturbation, slow perturbation, and control) consisted of 10 elbow extensions before, 30 elbow extensions receiving, and 10 elbow extensions following a pulling force at the participant's hand. The control condition did not consist of a pulling force during the middle 30 elbow extension trials.

All force conditions were implemented via a modified HuMoTech Caplex system (HuMoTech, Pittsburgh, PA) with a defined rotational speed of 12.48rad/sec. The forces applied to the participant were measured via a loadcell (LRF350, FUTEK Advanced Sensor Technology, Inc., Irvine CA) but not controlled other than software and hardware safety limits. The perturbation conditions consisted of a pull opposing elbow extension for 0.1sec.

Reaction time was considered the duration between cue onset and the initial muscle activation. Initial muscle activation was considered 2% above the maximum resting squared muscle activation.

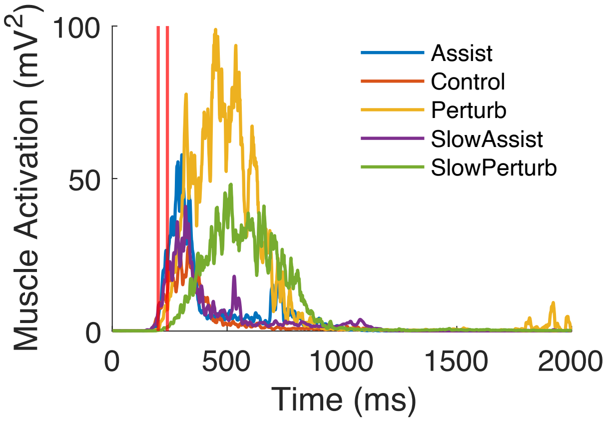

Figure 1: Average muscle activation across 30 lateral elbow extension trials for five different conditions. 'Assist' and 'SlowAssist' correspond to assistance ~20ms faster or ~20ms slower than muscle force production onset time respectively. 'Perturb' and 'SlowPerturb' correspond to a perturbation ~20ms faster or ~20ms slower than muscle force production onset time respectively. A moving average with window size of 10 frames (1000Hz sampling) was applied to the average muscle activation across trials for this figure only. The first vertical red line approximates timing of initial muscle activation at 200ms and the second vertical red line estimates the start of muscle contraction.

Results and Discussion

Changes in reaction time between the 10 extensions prior and 10 extension following force conditions were evaluated. assistance: 22.44ms, slow assistance: 8.50ms, perturbation: 0.04ms, slow perturbation: 9.55ms, and control: -12.22ms. However, the baseline for the control condition was significantly different from the other condition baselines ($P<0.05$). This current limitation will most likely be remedied with the addition of participants.

These results present preliminary evidence that small adjustments in the timing of application of force can result in differences in muscle activation patterns (Figure 1). Both assistance and perturbation force conditions with onsets prior to muscle contraction, or prior to muscle force production, illicit larger subsequent muscle activation to complete an elbow extension task.

Significance

These preliminary findings suggest that there may be sensitivity to the timing of force applied and therefore, may point to a greater need of consideration during robotic interventions. Furthermore, these findings present foundations for future investigations involving longer training paradigms geared to reducing reaction time in older adults and clinical populations with the intent of improving myelin formation.

Acknowledgments

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CLUTCH CATCH: THE PROSTHETIC BASEBALL GLOVE

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Introduction

Currently, upper-limb amputee baseball players have few reliable, user-friendly, cost-effective prostheses to compete with their able-bodied peers. In an ideal world, upper-limb amputees would be given equal opportunity to participate in activities that bring them joy. The lack of accessible resources for amputee athletes force many to quit the sport and results in decreased self-confidence and quality of life. To address this issue, we developed a functioning prototype with full catch-release glove mechanics to increase the accessibility and functionality of a baseball prosthetic. Through the testing of different springs, we optimized the spring type, size, and coiling to ensure our mechanical model will close a glove to securely catch a baseball.

Methods

The Clutch Catch is designed to mimic the human hand inside of a baseball glove, consisting of a rotating thumb rod, a fixed pinkie rod, a palm piece, and a wrist attachment to a residual limb. The prosthetic device is mechanically driven, with a torsional spring inside the palm that forces the thumb rod closed. Connected to the thumb rod is a slider piece that follows a slot cut into the outer surface of the palm. At the end of this slot is a notch, and to open the device, the thumb rod is pulled into position and the slider sits within the notch. This keeps the glove open until a baseball hits the glove, pushes the thumb rod and slider back out of the notch, and the thumb rod closes due to the spring force. The positioning of the thumb rod and pinkie rod when closed gives the glove plenty of closing to secure the baseball without it falling out, while also enough space for the player to reach in and grab the ball. This simple, automatic design allows the user the freedom of having their other hand to throw the baseball after catching, and the ability to reopen the glove in a timely manner.

Figure 1: *Clutch Catch assistive device. Clockwise from top left, slider view, interior view, device in glove.*

Results and Discussion

Currently, the market for hand amputees interested in competitive baseball is limited to a lacrosse stick head look-a-like without catch-release mechanics or the need for a baseball mitt. While designing the assistive device, the team identified the importance of the baseball player having his own mitt as it is a symbol of the game of baseball. The Clutch Catch fits inside of a regulation glove for high schoolers and college baseball athletes of 13 inches long and 8 inches wide.

We tested the reliability and consistency of the device under simulated game conditions, using a rating system of 1 and 0. A score of 1 represented a successful trial where the device securely caught the baseball, and 0 meant the device failed at catching the baseball. We used a total of 10 trials executed at each of the speed ranges a baseball mitt regularly experiences during a competitive game (30-39 MPH, 40-49 MPH, and 50-59 MPH). The Clutch Catch performed with a 100% success rate for speed ranges 30-39 MPH and 50-59 MPH while the success rate at 40-49 MPH was 90%.

We found no significant difference between the average rating in each speed range (t-test in Excel). The Clutch Catch performs consistently within these speed ranges. The prototype, which uses a PVC plastic cap for the palm, is rated to withstand up to 63 MPH. At this speed, the plastic palm broke in testing, but utilizing an aluminum palm would substantially increase the maximum speed rating. The prototype's thumb, pinkie, pin, and slider are all made from the same aluminum material, and after examination at the end of each testing trial, the parts experienced no physical damage, which confirmed that the Clutch Catch, once fully built from aluminum, could across a full season without damage.

Significance

Clutch Catch improves hand, wrist, and trans-radial arm amputees' quality of life by giving them the opportunity to play baseball competitively amongst their peers. Adoption of the Clutch Catch device would allow players to confidently participate in games by relying on the mechanics of the device for catching and easy release of the ball. The device allows the user to look and feel like their peers with the natural motions and the use of a glove.

Acknowledgments

Engineering Department at Elizabethtown College
The Miracle League
Friends and Family

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SENSITIVITY OF TRANSIENT BALANCE METRICS TO STIMULUS SYNCHRONIZATION

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Introduction

Transient periods (5-15 seconds) of increased postural sway following sensory transitions (e.g., closing eyes) have been reported during quiet stance balance control, which has spurred their potential association with sensory reweighting [1]. An epoch-based approach to quantify these transient features has proven insightful for sensory **(Figure 1)**, cognitive, and stance perturbations, and has shown discriminative ability for group differences in age and expertise [1-3]. However, the sensitivity of the epoch-based approach to differences in protocols remains unknown. As some balance protocols may allow participants time to adapt to a sensory-deprived condition (e.g., closing one's eyes) before starting to collect balance data, we sought to understand the impact of variability in this timing on epoch-based transient balance metrics. We hypothesized that a transient balance metric (**ΔEA**) would become less correlated as the time delay between eyes closure and trial initiation increased.

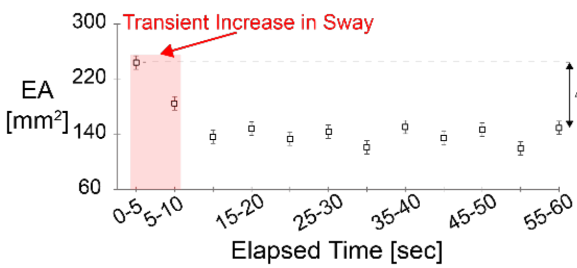

Figure 1: Ellipse area (EA) calculated for 5-second epochs throughout a 60-second quiet stance trials immediately following eyes closure [1].

Methods

Previously-collected eyes-closed balance data for 14 healthy, younger adults (23.9 ± 2.6 years, 79.8 ± 9.2 kg, 1.81 ± 0.07 m, 11 males/3 females) were re-analyzed for this study. Balance data consisted of center of pressure (CoP) recorded for a total of 60 seconds per trial. Trials were initiated by participants counting down '3-2-1-Go'. When participants said 'Go', they closed their eyes and a researcher started the 60-second data collection. Averages of three trials were used in analyses.

For this sensitivity analysis, a sliding 55-second portion of the trial was analyzed that aimed to emulate progressive amounts of time delay (0-5 seconds) between participants closing their eyes and the start of balance data. For example, 0-55 seconds was analyzed for the synchronized (unaltered) data, while 5-60 seconds of the trial was analyzed to artificially introduce a 5-second delay from closing eyes to starting CoP data collection. 95% confidence ellipse area (EA) was calculated for each epoch throughout the selected portion of the trial [1]. The difference in EA from the first epoch to the last epoch was used to characterize transient postural control behavior (**ΔEA - Figure 1**). Epoch lengths of 1, 2, 4, 6, and 10 seconds were used to gain insight into how the sensitivity to time delay may be influenced by epoch length. Spearman correlations for the transient measure (**ΔEA**) from the synchronized to the various time-delayed scenarios was the primary outcome for this study. We hypothesized that even short amounts of time delay would result in poorly correlated **ΔEA**, with the effects being more pronounced with short epochs.

Results and Discussion

Increased time elapsed between participants closing their eyes relative to the start of CoP data was generally associated with decreased correlations in **ΔEA** calculated from the same balance trials **(Figure 2).** Large decreases in correlation coefficients were seen after just 1-second of an artificially-induced time delay when using short epochs (e.g., 1-2 second epochs). Longer epoch lengths (e.g., ≥4 seconds) had more gradual decreases in correlation strength with increasing time delays, with these epoch lengths being largely robust to slight time delays. For example, **ΔEA** calculated with vs. without an artificially induced 1-second time delay was highly correlated (rho >0.87) when using epoch lengths of 4, 6, or 10 seconds.

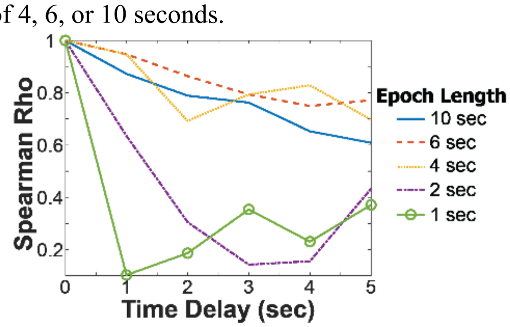

Figure 2: Spearman correlation coefficients for ΔEA calculated using center of pressure data with progressive amounts of artificially-induced time delays. Correlations are relative to ΔEA calculated for the synchronized (unaltered) data for a given epoch length.

Significance

These findings suggest that highly-consistent estimates of transient balance metrics are possible in the presence of small time delays between sensory transition and initiating balance data collection as long as moderate epoch lengths are used (i.e., 4-10 seconds). This finding may support the retrospective analysis of balance data using the proposed epoch-based approach even if protocols provided a brief period for individuals to settle into a balance condition (e.g., after closing their eyes), as long as the epoch length is sufficiently long. These findings ultimately need to be considered alongside future efforts to determine optimal values for epoch length for sensitivity and reliability of transient balance metrics.

Acknowledgements

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CHANGES IN BALANCE CONTROL DURING DUAL-TASKING AND PLANTAR TEMPERATURE INTERVENTION

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Introduction

Maintaining balance is more challenging under dual-task conditions, where motor and cognitive resources are used simultaneously [1]. Reducing sensory inputs also increases balance demand [2]. In contrast, increasing foot temperatures leads to improved balance performance [3]. However, no scientific investigations have combined both conditions. Hence, we hypothesized that decreased balance control during dual-tasking can be compensated through plantar warming.

Methods

Balance control of 33 young healthy adults (mean± SD: 23.1 ± 2.6 yrs, 15 ♂, 18 ♀) was measured pre and post plantar warming. The protocol consisted of two trials of 15 s for each of eight conditions (randomized): double and single leg stance with eyes open and closed, and with and without dual-tasking. Plantar warming (10 min at 35 °C) was induced by a thermal platform. An adapted Stroop-test constituted the dual-task test. Using a force plate, measured center of pressure (COP) parameters were: COP total, anteroposterior and mediolateral COP length, and COP average velocity. Mean±SD of the COPs were calculated for statistical analyses (α=0.05).

Results and Discussion

Plantar warming was successfully induced (6.4°C), hence, plantar sensitivity was likely increased [4]. During double leg stance, some parameters exhibited significant balance improvements (table 1). This is in line with [3]. However, our improvements were only evident during dual-tasking. During single leg stance, plantar warming significantly improved all COP parameters. It seems that effects are only evident in more challenging balance conditions, e.g., single leg stance and with cognitive load. We cannot exclude potential learning effects due to the order of the protocol.

Regarding dual-tasking, the additional cognitive demand did not show effects in all conditions. Interestingly, for single leg stance with eyes open, the dual-task induced significant postural instability. On the other hand, double leg stance showed significant improvements in balance control when dual-tasking. Our data suggests that the extra cognitive task during bipedal tasks with eyes closed resulted in improved attention and subsequently better motor performance, supporting the posture-first principle [5]. However, this idea has its limitations. When the total attention capacity is exceeded, especially during more challenging tasks (single leg stance), there are negative consequences for the balance control. This agrees with the findings from [6].

Considering the combination of warming intervention and dual tasking, we reject our hypothesis, since the warming effects were insufficient to compensate the consequences of the cognitive load. Further research is necessary to investigate the relationship between balance-improving interventions and cognitive processes, also considering muscular analysis.

Significance

Regardless of plantar warming, our data show different scenarios in which the dual tasking led to improved but also impaired balance control. It is important to identify such situations in activities of daily living to apply various interventions, especially in the elderly. Finally, this knowledge might help to reduce the risk of falls.

Acknowledgments

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Table 1: Mean±SD of balance data. Total (COP Total), ML (ML length) and AP (AP length) in [mm], and Vel (COP vel) in [cm/s].

	Pre warming intervention				Post warming intervention			
	Total	**ML**	**AP**	**Vel**	**Total**	**ML**	**AP**	**Vel**
DEWO	111.4±159.5	67.4±109.8	82.6±117.4	1.38±1.07	101.0±176.0	58.1±117.8^{E}	76.9±133.0	1.31±1.18
DEW	111.0±199.5	65.5±131.9*	84.7±152.0	1.38±1.34	86.1±135.8	46.4±90.3*^{E}	68.2±103.1	1.22±0.91
DCWO	167.6±185.2	89.5±132.3	131.2±132.7	1.76±1.24	136.3±124.5^{F}	69.4±86.5	106.9±91.6^{G}	1.55±0.83^{H}
DCW	147.9±145.6*	80.9±100.4*	113.6±107.4*	1.63±0.98	121.2±108.2*^{F}	61.2±74.9*	96.4±80.0*^{G}	1.45±0.72^{H}
SEWO	391.9±282.3*^{A}	251.1±203.7*^{B}	261.8±192.4*^{C}	3.25±1.89*^{D}	306.5±178.4*^{I}	175.0±110.3*^{J}	222.6±135.1*	2.69±1.18*^{K}
SEW	395.5±200.9*^{A}	253.5±135.2*^{B}	265.2±141.0*^{C}	3.30±1.34*^{D}	349.7±260.7*^{I}	211.8±165.0*^{J}	244.6±198.9*	2.98±1.73*^{K}
SCWO	1451.3±1866.0*	927.6±1337.9*	971.2±1286.1*	10.32±12.45*	958.7±369.7*	571.5±189.5*	647.3±287.3*	7.04±2.45*
SCW	1290.1±1114.5*	814.2±791.8*	857.9±769.2*	9.25±7.43*	976.9±339.0*	594.7±179.9*	649.2±258.5*	7.16±2.25*

* $p<0.05$ of Wilcoxon test (pre vs. post comparisons) and $^{A, B, C, D, E, F, G, H, I, J, \text{and } K}$ $p<0.05$ of Wilcoxon test (with vs. without dual-task). D double leg, S Single leg, E eyes open, C eyes closed, W with WO without dual-tasking.

THE IMPACTS OF MENTAL FATIGUE ON BALANCE CONTROL

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Introduction

Mental fatigue (MF) is a psychobiological state experienced following a period of prolonged cognitive exertion[1,6]. Common symptoms of mental fatigue include general feelings of tiredness or lack of energy.[1,6] A recent meta-analysis[2] suggested that MF can reduce aspects of aerobic-, motor control-, dynamic resistance-, and isometric resistance-based task performance. Recently, literature has begun to examine the impacts of MF on aspects of balance-based task performance, with some research indicating an increased likelihood of loss of balance and slip-induced falls after a performing a task to induce MF.[5,7] While another recent study[3] found that MF only increased postural sway in older adults and not younger adults. However, little research has been conducted examining the impact of MF on postural stability during static standing tasks.

As such, the purpose of this study is to examine the impact of MF on balance performance during static standing trials. We hypothesized that mental fatigue would lead to an increase in postural sway resulting from a decrease in postural stability.

Methods

27 participants were recruited for this study and were randomized either into an experimental (n=13) or control (n=14) group. All participants were healthy adults between the ages of 18 and 28. Participants performed two postural sway trials separated by the performance of a cognitive task. During the postural sway trials, participants stood on an AMTI force platform (AccuGait 300lb max range, AMTI) for 15 seconds. During these trials, participants were asked to stand as still as possible.

Participants in the experimental group were asked to perform a 10-minute, incongruent Stroop task[8] between sway trials. Participants in the control group were asked to watch a 10-minute clip of an emotionally neutral documentary (i.e., Our Planet) between sway trials. Effectiveness of the mental fatigue manipulation was monitored using a mental fatigue visual analogue scale (MF-VAS) in which participants were asked to indicate their perception of their current level of mental fatigue on a visual analogue scale (VAS) from 0 (none at all) to 100 (maximal). Participants completed the MF-VAS scale a total of 6 times, once before commencing the 10-minute cognitive task and at 2-minute intervals throughout the task. MF-VAS scores were calculated by summing the scores across the 6 measurements. Mean amplitude (MA) of centre of pressure position (CoP) was calculated to provide an estimate of postural stability. MA represents the average distance of the participant's CoP from their mean CoP location.[4]

Results and Discussion

MF-VAS scores were significantly higher in the MF group when compared to the control group ($p < 0.001$). This suggests that the Stroop task was successful at inducing MF.

Results of a repeated measures ANOVA showed no significant effect of MF on MA between conditions ($p = .901$). This suggests that there may be negligible carryover effects of MF on postural stability during static standing tasks. As such, future research is needed to determine the potential impact of MF on postural stability during dynamic tasks.

Table 1: Balance and mental fatigue data summary.
MF-VAS, Mental fatigue visual analog scale; MA, mean center of pressure amplitude. Mean(SD) data shown

Measured Outcomes	Mental Fatigue	Control
MF-VAS	310.5 (95.4)	119.6 (84.4)
MA		
Pre-10 min task	50.06 (27.61)	55.02 (19.16)
Post-10 min task	44.18 (22.19)	50.39 (25.11)

Significance

By studying biomechanical outcomes such as balance, this may help elucidate the mechanisms upon which mental fatigue are acting. It may be especially important to monitor the impact of mental fatigue on balance for individuals who are at risk of fall-based injuries[9]. In turn, this could provide clinicians a better understanding of the potential impact that mental fatigue has on balance and allow for the implementation of fall preventative measures to reduce fall-based injuries.

Acknowledgments

This work was supported by NSERC Discovery Grants Program.

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Wearing a Backpack Affects Lower-Limb Muscle Activations During Balance Recovery by Stepping

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Introduction

Backpacks are a common form of loads that people carry. Wearing a backpack may increase fall risk more than other methods of load carriage, due to the dorsal placement of the load (Qu & Nussbaum, 2009). Carrying a load increases lower limb muscle activation during gait (Silder, 2013). However, how wearing a backpack affects lower-limb muscle activations when recovering balance is not known. Thus, we examined the influence of wearing a backpack on the amplitude of lower limb muscle activations. We hypothesized that wearing a backpack would increase mean and peak muscle activations during postural responses to both forward and backward lean-and-release perturbations.

Methods

Thirteen healthy, young adults, between the ages of 18 and 40 years participated in the study [7 female, 6 male; mean(SD) age: 24(2) y; stature: 1.7(0.1) m; mass: 74.6(12.3) kg]. Participants performed forward and backward lean-and-release balance recovery trials with and without wearing a backpack set to 15% of individual body weight. Participants started at release angles of 2º and 10º for backward and forward perturbations, respectively. The release angle was increased incrementally between trials by 2º until the participants could not recover their balance with only one step. Bilateral muscle activity of the tibialis anterior (TA), medial gastrocnemius (MG), rectus femoris (RF), and biceps femoris (BF) were recorded at 1500 Hz using an 8-channel wireless surface electromyography (EMG) system (Noraxon, USA). EMG data were de-meaned, bandpass filtered (20-450 Hz, fourth order, zero-lag, Butterworth), full-wave rectified, and the low-pass filtered (10 Hz cutoff, fourth-order, zero-lag, Butterworth). Muscle activations were normalized to peak RMS reached during initial maximum trials. Mean and maximum activations were calculated during step initiation: from tether release until toe-off. These activations were calculated for both the stepping and standing leg. Activations were compared between loaded and unloaded conditions at the greatest lean angle successfully recovered from, using paired t tests (significance concluded when $p < 0.05$).

Results and Discussion

Lean angles ranged from 14 to 36° in the forward lean direction (22(5)°), and 4 to 14° in the backward lean direction (10(3)°). Maximum lean angle was not significantly affected by load in either direction (forward: p=0.45, backward: p=0.09). Maximal MG activation in the stepping leg in backward leaning was significantly greater during loaded (51(25) %max) than unloaded (32(19) %max) conditions (p<0.01). Conversely, during backward perturbations, maximum activation of the standing leg TA was smaller while loaded (41(27) %max) than unloaded (59(16) %max). Excepting the TA and MG of the standing leg during backward leaning trials, peak muscle activations were greater when loaded. We found no significant differences in muscle activations between loaded and unloaded releases for the RF or BF (p>.05). There were also no significant differences in mean muscular activations between loaded and unloaded conditions.

These results suggest that the addition of a load may require the modification of lower limb muscle activations to accommodate the center-of-mass acceleration caused by the additional load on the trunk (Pitts et al., 2021). In backward perturbations, the increased MG activity may be required to produce the larger stepping response we reported earlier (Pitts et al., 2021).

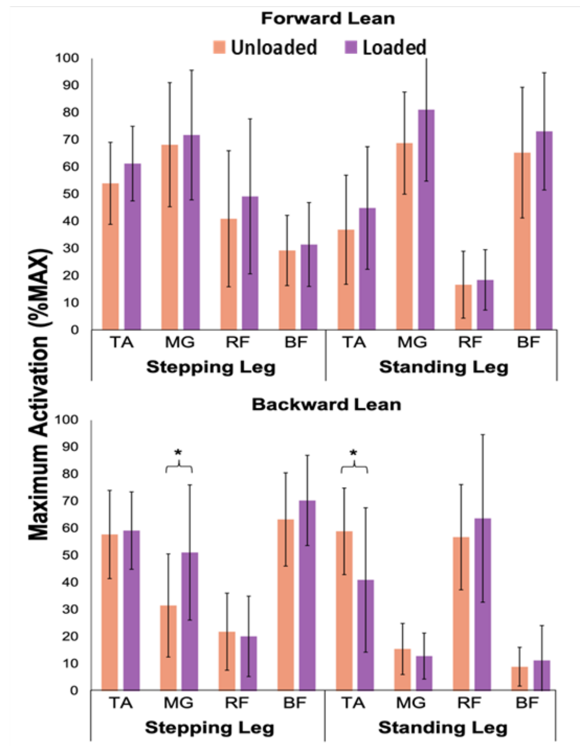

Figure 1: Maximum muscle activation during forward and backward lean perturbations (* = p<0.05). Bars and whiskers represent mean and standard deviation respectively.

Significance

Our findings provide further evidence of the neuromechanical effects of backpacks on reactive balance control. Increases in lower limb muscle activations did not affect the ability of young and healthy adults to recover balance. However, the required adaptations may pose a challenge for other populations that lack the ability to effectively generate greater muscle activations in response to the increased center-of-mass acceleration of the additional load.

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Reliability and validity of the angle measurements of inertial measurement unit sensors in headphone and necklace posture correction system for office workers

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Introduction

Office workers experience neck or back pain due to the poor posture that occurs during long-term sedentary work [1]. Posture correction is used to reduce pain caused by poor posture and ensures proper alignment of the body. Assistive devices have been developed to assist in maintaining an ideal posture, but there are limitations in many situations. We developed headphone and necklace system for posture correction with inertial measurement unit (IMU) sensor that gives visual or auditory feedback. We have evaluated the angular measurement reliability and validity of IMU sensor compared with three-dimensional motion analysis system (Vicon Motion system, Vicon, Oxford, UK).

Method

Seven young adults (six males and one female) aged between 18 and 35 participated. The study protocol was approved by the Institutional Review Board at Yonsei University Mirae Campus. Informed consent was obtained from all participants.

Participants wore headphone- and necklace- applied sensors. Markers from the Vicon were attached to the headband of the headphones and the sensor that placed on the upper back. They sat in a chair, maintained the hip and knee flexion at about 90 degrees, fixed the sole of their feet on the floor, and restricted their lumbar movements using belts. They practiced head-cervical flexion and upper thoracic flexion to familiarize themselves with the movement.

Sessions were conducted twice a day, repeating the head-cervical flexion and upper thoracic flexion three times. The sequence of actions was randomized. A rest of 30 minutes was provided between sessions. The flexion angle for each action was measured simultaneously by posture correction system and Vicon.

The data of posture correction system and Vicon were analyzed through the intra-class correlation coefficient (ICC [3,3]) with 95% confidence intervals (CIs) using the two-way mixed ANOVA average measure and Pearson's correlation coefficient (r).

Results and Discussion

We found that the measurement for head-cervical flexion and upper thoracic flexion had high intra-rater test-retest reliability using the posture correction system and Vicon with ICC, standard error of measurement (SEM), and minimal detectable change (MDC) values (Table 1). The angle measurements of the system were highly correlated with those of the Vicon in head-cervical flexion (r=0.995) and upper thoracic flexion (r=0.951) in session 1. The angle measurements of the system also showed high correlation with those of Vicon in head-cervical flexion (r=0.978) and upper thoracic flexion (r=0.912) in session 2. The scatter plots show the relationship between the IMU sensor in the system and Vicon (Figure 1).

Previous studies have also shown high reliability and validity for the measurement of the angle of an IMU sensor [2,3]. Our findings showed higher reliability and validity than those of the previous studies. Unlike previous studies, we used belts to minimize unwanted lumbar and trunk movements in sitting positions during head-cervical flexion and upper thoracic flexion. In addition, no studies have been conducted on IMU sensors that provide auditory and visual feedback through headphones and monitors.

Table 1. Reliability of the flexion angle measurement

	IMU Sensor			Vicon		
	ICC (95% CI[a])	SEM[b]	MDC[c]	ICC (95% CI[a])	SEM[b]	MDC[c]
C[d]	0.97 (0.86-0.99)	1.37	3.79	0.97 (0.82-0.99)	1.57	4.35
T[e]	0.95 (0.72-0.99)	1.55	4.29	0.95 (0.72-0.99)	1.45	4.01

[a]confidence intervals, [b]standard error of measurement, [c]minimal detectable change, [d]cervical flexion, [e]thoracic flexion

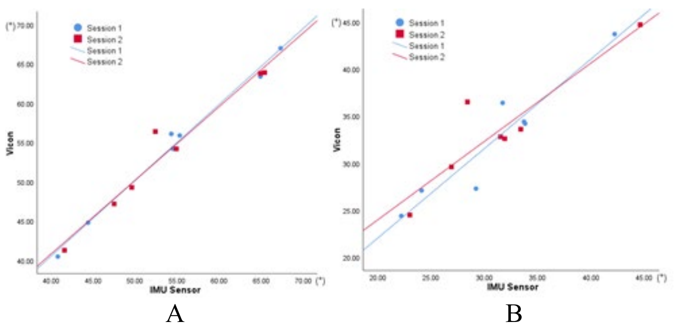

Figure 1: Scatterplot of head-cervical flexion (A) and upper thoracic flexion (B) angle measurements

Significance

Our study demonstrates that headphone and necklace posture correction system is as reliable as the Vicon in measuring angle compared with Vicon, the gold standard. IMU Sensors can accurately measure the angle changes and provide feedback on time. The system has less setting work than the Vicon and is easy to apply in the actual working environment. Therefore, headphones and necklace posture correction system would be useful for office workers to correct their poor posture that occurs during long-term sedentary work.

Acknowledgement

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SEGMENTAL FRONTAL PLANE ANGULAR MOMENTUM DURING PRE-PLANNED AND LATE-CUED TURNS

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Introduction

Depending on the environment, turning can account for up to 50% of all walking steps [1]. However, turning can be more challenging than straight-line gait. In fact, 70% of falls during turns result in injury [2]. Furthermore, turning with a late cue (LC), could challenge balance more compared to a pre-planned turn (PP), when the cue to turn is perceived early with respect to movement execution. Whole-body angular momentum (H) about the center of mass (COM), which is the sum of each segment's angular momentum, is a common measure used to assess dynamic balance. Unregulated angular momentum can lead to a loss of balance and falls, especially in the frontal plane, as it is often challenging for individuals with neuromuscular impairments to control balance in the frontal plane during straight-line gait [3]. The purpose of this study is to understand how balance is controlled at the segment level during PP and LC turns. We hypothesize that when turning with a late cue, the range of the frontal plane angular momentum (Hf) of the legs, torso, and pelvis will be greater compared to the pre-planned turn.

Methods

Ten healthy young adults, all of whom were free of pathologies and pain that would have affected them during daily walking, participated in this study (3 f, 7 m; 25.2 ± 4.2 yrs.; 73.9 ± 14.8 kg; 1.79 ± 0.1 m) after providing informed consent in accordance with the IRB. A 15-segment whole-body kinematic model [4] was created using optical motion capture (200 fps; Optitrack, USA). Participants were instructed to walk as though they were in the grocery store and not in a rush. They performed 10 PP and 10 LC 90º turns to the left after walking 5 m. During LC turns, the participant approached the turning intersection and viewed an 84" monitor at the end of perpendicular aisle for either a sign to turn, or a red "NO" symbol to continue walking straight. All analyses were performed in MATLAB (Mathworks, USA). Left and right leg Hf were computed by summing left and right foot, shank, and thigh Hf, respectively. Negative Hf indicated rotation of the body above the COM leftward towards the direction of the turn.

Turns began at the heel strike prior to pelvis heading angles increasing three standard deviations above person-specific straight-line gait values. Turns ended at the first heel strike after pelvis heading angles decreased below that same threshold, with respect to the new direction of travel. During the turning phase, we computed the maximum, minimum, and range of Hf for the outside leg, inside leg, torso, pelvis, and whole body. For every subject, the median within each metric was found across 10 trials for each LC and PP. A paired sign test (α=0.05) was used to compare each metric's median between turn types for each segment and the whole-body.

Results and Discussion

Hf ranges and minima are provided in **Fig. 1**. During LC turns, there were significantly greater ranges of Hf vs. PP turns for the torso (p=0.022) and outside leg (p=0.002) (**Fig. 1B**). This is driven by significantly smaller minima Hf of the outside leg and torso during LC vs. PP turns (each p=0.02) (**Fig. 1A**). However, there were no significant differences in Hf range for the inside leg (p=0.75), pelvis (p=0.34), and whole-body (p=0.11) across turn types. There were no significant differences in Hf maxima for any segment or for the whole-body across turns.

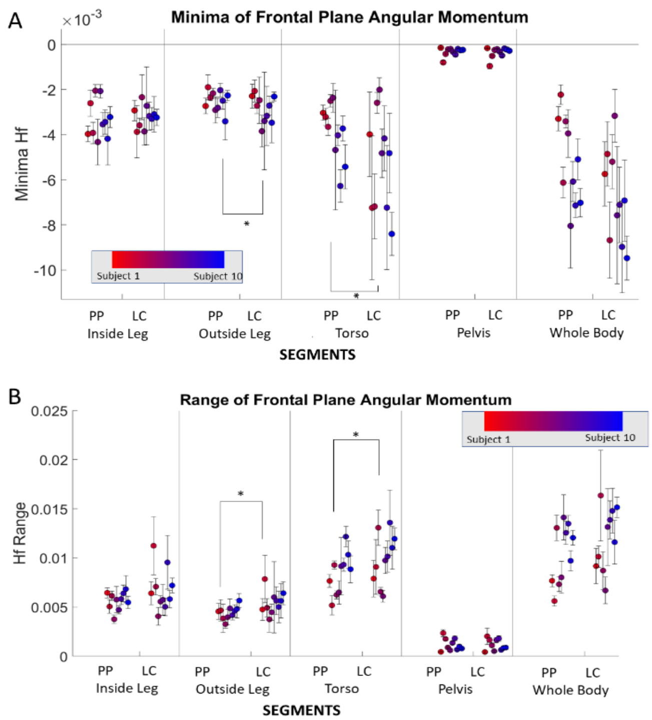

Figure 1. Median (standard deviation) (**A**) Hf minima and (**B**) Hf range of body segments and the whole-body during pre-planned (PP) vs. late-cued (LC) turns for each subject (1-10). Brackets indicate group-level significant differences between turn types (p<0.05).

Overall, we found that the outside leg and the torso use smaller minima Hf and thus, greater ranges of Hf during LC vs. PP turns. In the future, we will analyze the gait event contexts for these significantly different minima during PP and LC turns. Additionally, including more participants could provide more insight for group-level comparisons given the observed across-participant variability in segmental and whole-body Hf.

Significance

Our findings of more extreme Hf minima and larger Hf ranges for the outside leg and torso during LC vs. PP turns provides initial evidence that LC turns may require different angular momenta control patterns across body segments than do PP turns. Understanding more about the relationship between segmental and whole-body angular momenta during different turn contexts could assist future balance training and rehabilitation programs.

Acknowledgments

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MOTOR ADAPTATION IN STEP WIDTH CONTROL FOLLOWING SYSTEMATIC MEDIOLATERAL PERTURBATIONS TO FOOT PLACEMENT: PRELIMINARY STUDY

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Introduction

To walk without falling, steps should be placed in the appropriate position[1]. The body's center of mass must be controlled by stepping feet to support dynamic balance on each step. Many studies have shown that maintaining balance is more challenging in the frontal plane than in the sagittal plane, requiring active mediolateral balance control[2]. Mediolateral foot placement control in various conditions should be examined.

Humans learn locomotion while interacting in various environments during their lifetime. As sensorimotor relationships are systematically modified by different body or environmental contexts, the motor system experiences unexpected sensory feedback and subsequent motions are quickly modified to compensate for the changes[3]. Motor adaptation is likely crucial for balance maintenance, but we do not know if people use step width adaptation to control frontal plane balance.

Our study aims to determine the biomechanical responses during and after repeated medial or lateral perturbations to foot placement. We hypothesize that, following these systematic perturbations, people will exhibit error-based motor control to modify their step width, and that they will maintain a step width-based balance control.

Methods

Here we report preliminary data from two healthy adult participants (one male and one female, ages = 28, 27). The participants walked in the Computer Assisted Rehabilitation Environment (Motek; Amsterdam, Netherlands), consisting of a treadmill mounted on a 6-degree-of-freedom motion platform. Ten infrared cameras tracked retroreflective markers attached to the participant's pelvis and feet. Each session consisted of five walking periods (Figure 1). The 1st, 3rd, and 5th were normal walking periods (NW1, NW2, and NW3) of 10 minutes, while the 2nd and 4th were perturbation periods (P1 and P2) of 5 minutes. During perturbation periods, D-Flow software recognized gait phases using kinematic data and made 5 cm of medial or lateral platform translations right after every toe-off. For participant 1, lateral perturbations were administered in the 2nd period, and medial perturbations were administered in the 4th period. For participant 2, the order of perturbations periods was reversed.

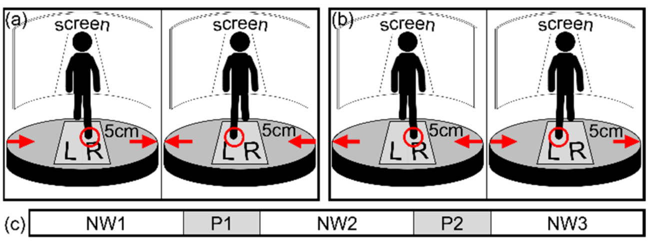
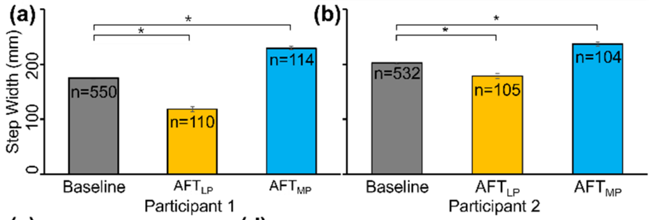


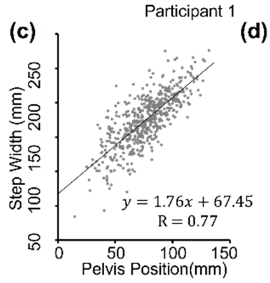
Figure 1: lateral perturbations (a) and medial perturbations (b) administered while walking. Perturbation periods were preceded by and interspersed with periods of normal walking according to the order of periods (c).

Results and Discussion

Testing consisted of a baseline condition and after effect conditions following lateral perturbations (AFT_{LP}) and medial perturbations (AFT_{MP}). Baseline involved the last 5 minutes of the first normal walking period, and after effect conditions involved the early 1 minute of normal walking sections right after the perturbation period. The average step width of AFT_{LP} was significantly reduced compared to baseline for each subject (Figure 2a,b; subject 1: $t(658)=13.23$, $p<0.001$; subject 2: $t(635)=6.01$, $p<0.001$), while the average step width of AFT_{MP} significantly increased compared to baseline for each subject (subject 1: $t(662)=14.08$, $p<0.001$; subject 2: $t(634)=8.89$, $p<0.001$).

Wang et al.[4] reported that each step width can be predicted by the position of pelvis at the preceding midstance. For example, at left midstance, rightward pelvic deviation result in a wider step width in next right step with a linear gain. We used linear regression to determine the association between step widths and preceding pelvic positions. Significant associations were shown for all conditions (figure 2d). Based on the gain, (the ratio of the step width deviation to pelvic deviation), step width was controlled according to the preceding position of the pelvis.





(d)

		Step Width ± SD (mm)	Gain	R	p
Participant 1	Baseline	175±38	1.76	0.77	< 0.001
	AFT$_{LP}$	118±47	1.93	0.74	< 0.001
	AFT$_{MP}$	229±33	1.85	0.76	< 0.001
Participant 2	Baseline	202±35	1.86	0.80	< 0.001
	AFT$_{LP}$	178±44	1.74	0.74	< 0.001
	AFT$_{MP}$	237±40	1.47	0.62	< 0.001

Figure 2: The average step width and standard error is shown for different conditions of participant 1 (a) and participant 2 (b). An example of linear regression between pelvic position at midstance and step width from baseline walking of participant 1 (c). Table (d) represents the gain and strength of association.

Significance

Preliminary data indicates that, during the AFT periods, the relationship between pelvic position and step width is actively maintained despite modified step width in these conditions. Further research will be conducted to test these hypotheses with a larger sample and determine the biomechanical mechanisms for modulating step placement.

Acknowledgments

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DOES LOCOMOTOR REACTION TIME GENERALIZE BETWEEN GAIT INITIATION AND WALKING?

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Introduction

Falls are a persistent public health concern, and most falls occur during locomotor activities such as walking. The ability to quickly respond to unexpected balance challenges while walking is essential for mitigating fall risk. However, walking is a complex task for which measuring locomotor reaction times relevant to instability and fall risks can be challenging. Potentially for that reason, previous research has commonly measured reaction time during gait initiation as a more feasible proxy [1-3]. Indeed, inferences from gait initiation often extend to walking. However, recent studies suggest that the coordination of balance control mechanisms can vary between tasks [4]. Unfortunately, it is currently unclear if locomotor reaction time generalizes between gait initiation and response to unexpected balance challenges during walking. Thus, the purpose of this study was to evaluate the extent to which reaction time during gait initiation may serve as a proxy for that during a walking task known to challenge walking balance – in this case, lateral precision stepping. We hypothesized that reaction times during gait initiation would positively correlate with those when performing a lateral precision stepping task during walking.

Methods

15 healthy young adults (21.3 ± 2.18 yrs, 6 F) completed a gait initiation task 5 times bilaterally in randomized order. Subjects were uninformed of the trial step limb prior to a "Left" or "Right" verbal cue, and reaction times were calculated as the time between limb indication and the instant of toe-off. Subjects also performed two treadmill walking tasks at their preferred overground walking speed: a targeted lateral precision stepping trial and two minutes of normal walking. During the precision stepping trial [5], 4 laser targets (either 26 cm or 38 cm from midline) were projected onto the treadmill surface 5 times each per side. Targets were displayed in randomized order upon ipsilateral heel strike and disappeared after the targeted stepping attempt. Reaction times for precision stepping trials were calculated from target projection (heel-strike) to the instant of lateral swing limb deviation (i.e., +2SD from usual swing trajectory [5]). Force data were collected for all trials and subjects wore 36 motion capture markers on their trunk, pelvis, legs, and feet. We calculated Pearson correlations bilaterally and unilaterally for reaction time between tasks.

Results and Discussion

Group-average reaction time for gait initiation was 0.56 ± 0.13 s (0.58 ± 0.14 s for left limb initiation and 0.54 ± 0.13 s for right limb initiation). Group-average reaction time for the precision stepping task was 0.85 ± 0.06 s (0.85 ± 0.06 s for left targets, and 0.84 ± 0.07 s for right targets). In contrast to our hypothesis, we found no significant or meaningful correlations between reaction times during gait initiation and those during the precision stepping reaction task (Fig. 1A-C). This outcome held whether the correlations were performed for the bilateral average (r=0.118, p=0.665) or ipsilaterally for the left (r=0.167, p=0.538) or right (r=0.227, p=0.398) limbs, separately. We interpret this lack of correlation as evidence that reaction time during gait initiation is independent from that during walking because it is: (i) governed by distinct balance control mechanisms, (ii) subject to fewer and less immediate task constraints, and/or (iii) differentially sensitive to demographic or neuropsychological factors. Consistent with these latter two possibilities, we also observed that inter-subject variance (i.e., standard deviation) for gait intiation reaction time was notably larger than than that for precision stepping reaction time.

Significance

Our results suggest that reaction time does not generalize between gait initiation and precision stepping during walking. Although additional analysis of these data and future work is needed, we caution against using reaction times assessed during gait initiation as a proxy for those actualized during walking tasks relevant to balance performance.

Acknowledgments

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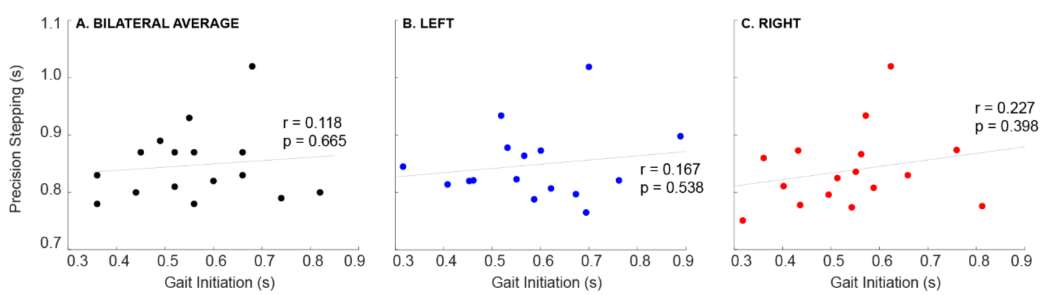

Figure 1. Univariate linear regressions between reaction time during gait initiation and that during precision stepping in walking.

MODEL ANALYSIS OF ABNORMAL FOOT PLACEMENT IN PEOPLE WITH VESTIBULAR HYPOFUNCTION

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Introduction

People with vestibular hypofunction (PwVH) often have unstable gait and increased fall risk [1]. Regulating foot placement (FP) is the primary strategy for gait stability [2]. Previous work has shown that PwVH demonstrate increased step width and step width variability [3,4]. In response to mechanical perturbations, PwVH show increased magnitude and variability in FP responses as well as take more steps to recover compared to healthy controls [5,6]. Potential explanations for these different responses to perturbations include: increased sensory noise, decreased sensitivity to motion, and increased lateral stepping in steady state walking to compensate for uncertain estimates [5]. The purpose of this work is to use a simple gait model to determine which of these mechanisms underlies the abnormal FP responses in PwVH.

Methods

A computational model was developed, consisting of a three-dimensional rigid body representative of a head-arms-trunk segment moved by massless legs. Foot placements were chosen to maintain stable walking with specified forward and lateral center of mass (COM) velocities based on linear inverted pendulum dynamics [7]. Only single leg stance was considered. The head-arms-trunk segment was limited to one degree of freedom in roll, as this is the plane in which PwVH show abnormal trunk sway [3,5]. The head-arms-trunk posture was maintained upright by a proportional-derivative controller. Segment mass and lengths were based on realistic anthropometric values. The experimental protocol described in [5] was replicated in simulation, where a 7cm FP perturbation was applied after a right heel strike in either a forward right or backward left direction. FP responses to the perturbations were quantified as the mediolateral (ML) moment arm between the foot position and COM. Four different model versions were developed and simulated for ten trials each: 1. model with increased sensory noise to COM velocity and rotational states of head-arms-trunk segment; 2. model with decreased sensitivity to rotational states of head-arms-trunk segment; 3. model with increased lateral stepping; and 4. model representative of the healthy state.

Results and Discussion

FP responses for the healthy version of the model closely matched experimental trends from [5], where forward right perturbations lead to wider recovery steps and backward left perturbations lead to narrower recovery steps. This illustrates that the model captures human-like stepping behavior. ML moment arms sampled at midstance are shown in Figure 1 for a forward right perturbation. The perturbation occurred on the fifth step (right), indicated by the first filled black circle, and the primary response occurs on the next left step. Increased sensory noise resulted in increased variability of FP response. The increased lateral stepping strategy led to an increase in decay time for FP responses to return to baseline, as the next right step post-perturbation is different from the baseline. Decreasing sensitivity to rotational inputs had little effect on FP behavior. Overall, these results suggest that abnormal foot placement control in PwVH can be attributed to increased sensory noise and/or increased lateral stepping.

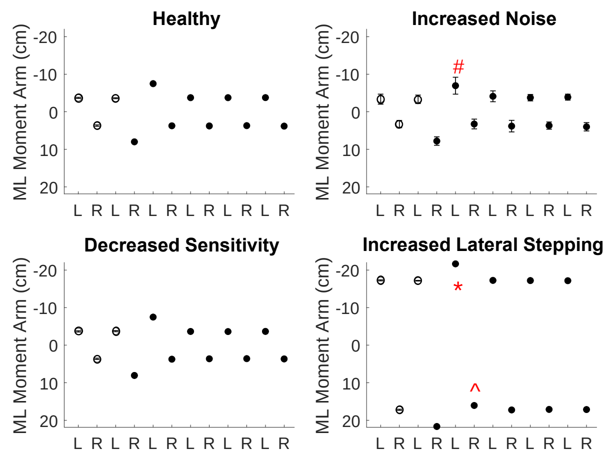

Figure 1: ML moment arms in response to forward right FP perturbation. Circles indicate steps; filled circles indicate steps during and after perturbation. Y-axis is reversed to visualize walking from left to right. #: increased variance compared to healthy; *: increased moment arm compared to healthy; ^: different moment arm from baseline value

Significance

Using a computational model allows for testing of specific hypotheses regarding the cause of abnormal FP control in PwVH that would be difficult to experimentally assess. The results here suggest that increased sensory noise, as opposed to purely decreased sensitivity, may be primarily responsible for the differences in FP responses. These results support using therapies focused on sensory reweighting to rely on "less noisy" sensory feedback (i.e. visual and proprioceptive input). Further modelling and confirmatory experimental work investigating other balance strategies in PwVH will improve the understanding of why this population has poor gait stability and possibly identify better treatments.

Acknowledgments

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CHANGES TO LOCOMOTOR PATH TRAJECTORIES FOLLOWING A VISUAL PERTURBATION

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Introduction

Vision allows us to safely navigate environments by choosing paths that provide us with more space and avoid obstacles [1]. Safe navigation is established by treating objects and goals in our environment as repellers and attractors respectively and we tend to not deviate much from our locomotor trajectory on route to a goal [2]. Vision, then, contributes to safe navigation by providing information about the rate and direction of self-motion relative to environmental features. The purpose of the current study was to determine whether a medial-lateral visual perturbation during locomotion would cause individuals to quickly realign their locomotor trajectory with their original path or choose a new one. It was hypothesized that participants' desire to realign to their original locomotor trajectory would be dependent on the magnitude of the visual perturbation and where the participant perceives the goal to be located. It was also expected that gait parameters (step length, SL; step width, SW) after the perturbations would reflect a re-established locomotor trajectory.

Methods

Participants (N=8, 21.13 years) were outfitted with an HTC Vive Pro head-mounted display unit to display the virtual world consisting of two poles and a goal. Participants were asked to walk along an 8-meter pathway towards a goal and avoid colliding with obstacles (i.e., 2 poles creating a 1m aperture) located 5m from the starting position. Participants wore 3 Optotrak rigid bodies located on the trunk and the distal ends of both legs with a sample frequency of 60Hz. Additionally, digitized imaginary points on the left and right glenohumeral joints and ASIS were used to estimate the location of the trunk COM. On a subset of trials (i.e., 30%), the visual world was perturbed 10, 20, or 30 degrees within 250ms, leftward or rightward when participants were 2m from the aperture to determine if and how a visual perturbation affects their physical world locomotor trajectory. For instance, the 10° shift kept the participants within the new virtual aperture boundaries, the 20° shift put the participants in line with one of the poles, and the 30° shift visually placed participants outside the aperture.

Results and Discussion

Kinematic data (ML COM position) at the time of passing the aperture was used to quantify participants' path selection. Preliminary results revealed that participants were motivated to pass through the aperture with virtual perturbations of 10° and 20°off their locomotor trajectory on 96% and 75% of trials respectively. However, when the virtual world was perturbed 30°, participants adjusted their pathway to walk through the aperture 23% of the time.

SL and SW for the step after the visual perturbation were quantified to determine whether there was an effect of perturbation magnitude. One-way repeated measures ANOVAs were run to evaluate both gait parameters. Results revealed that SL was significantly shorter for the 20° perturbation (59.8cm) in comparison to no perturbation (66.1cm) ($F_{(3,21)}$ =4.03, p=.02, η^2_p=0.37). However, SW was not significantly different between perturbation magnitudes ($F_{(3,21)}$ =2.81, p=.064) (Figure 1).

The current study's ML visual perturbations during goal-directed locomotion had three expected outcomes: 1) small virtual perturbations participants would realign their path with the new goal location and pass through the aperture because this would not change gait parameters significantly; 2) moderate virtual perturbations would result in participants realigning their locomotor trajectory to the new virtual goal approximately 50% of the time depending on foot in single support at the time of perturbation; and 3) large virtual perturbations would result in participants maintaining their original physical locomotor trajectory and not pass through the aperture on route to the virtual goal's new location because doing so would necessitate greater changes to gait parameters.

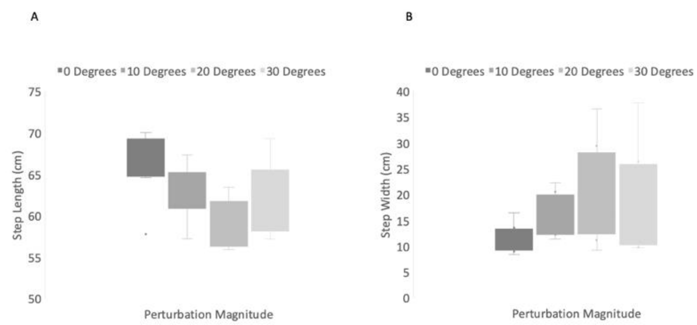

Figure 1: A) SL is significantly shorter following 20° perturbation than 0°. B) SW is not different across perturbation magnitudes.

It appears the attraction towards the middle of the aperture was strong enough to warrant passage with small (10°) or moderate (20°) visual perturbations on almost every trial. The gait kinematics during the moderate perturbations suggests that participants made significant adjustments (i.e., decreased SL and slight increase in SW) to their normal walking to steer towards and walk through the aperture on route to the new virtual goal location. In this case, vision may have a stronger influence on future locomotor trajectories/paths than gait biomechanics [3].

Contrary to our assumptions, the large visual perturbations were successful in eliciting changes in participants' trajectories (i.e., passing through the new aperture location) on 23% of the trials. Which suggests that the attraction to the middle of the aperture was stronger than minimizing biomechanical changes to gait parameters. Non-significant changes to gait parameters during the large perturbations could reflect the few instances in which individuals changed physical locomotor trajectories.

Significance

Virtual reality is an effective way of eliciting visual perturbations [4] that could be used in clinical settings. People experiencing attentionally demanding clinical deficits (i.e., pain, Parkinson's disease, etc.) may not be able to respond efficiently to visually delivered perturbations. An obstacle avoidance virtual reality perturbation paradigm could be used clinically as a means of testing for deficits in attention, and adaptive locomotion.

Acknowledgments

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EFFECTS OF PHYSICAL CERTAINTY OF DISCRETE UNDERFOOT PERTURBATIONS ON ANTICIPATORY AND REACTIVE BALANCE

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Introduction

Ambulation over complex terrain requires active control of foot placement to maintain a normal kinematic relationship between the center of mass and base of support[1]. When walking surfaces feature unpredictable underfoot disturbances individuals often adopt cautious gait strategies to maintain stability. However, recent investigations have suggested that foot placement location may be selected to mitigate predictable shifts to the underfoot center of pressure (CoP). However, those results relied on acceleration data, and specific strategies could not be characterized[2]. This study uses positional marker data to investigate anticipatory and reactive locomotor strategies for repeated underfoot perturbations with varying levels of physical certainty. We predicted that greater certainty about upcoming perturbations would result in participants adopting a perturbation-specific anticipatory strategy.

Methods

Thirteen healthy adults completed walking trials on an instrumented treadmill with random underfoot perturbations from a mechanized shoe. As part of a larger protocol, participants completed three five-minute walking trials, during which the left shoe randomly delivered an underfoot perturbation every 5-9 strides. Participants received a warning tone one stride before each perturbation to alert when a perturbation would occur. The physical nature of the perturbation varied: participants completed trials with 1) only eversion perturbations, 2) only inversion perturbations, or 3) randomly switching eversion and inversion perturbations (i.e., physical uncertainty). Kinetic and kinematic data were collected from an instrumented treadmill and retroreflective marker-based motion capture system. Step width (SW) and margin of stability (MoS) were calculated for each step. The percent change to SW and MoS, relative to the intra-trial non-perturbation steps, were the primary outcome measures. Linear mixed-effects models were fit for each step to identify differences between trials. A type III test of fixed effects and pairwise contrasts were used to assess significant differences between trials using an alpha of 0.05.

Results and Discussion

Physical certainty had little effect on SW or MoS in the steps leading up to the perturbation. For the step of the perturbation, MoS significantly differed between eversion and inversion trials ($p = 0.004$), likely due to the medial shift of the CoP during inversion and the lateral shift of the CoP for eversion relative to normal steps. Following the perturbation, participants exhibited a larger SW relative to normal steps for all conditions. Differences in SW relative to normal gait peak and return to normal more quickly in trials with only one perturbation type (high physical certainty) compared to the mixed trial (physical uncertainty) ($p < 0.001$). Specifically, SW for the inversion and eversion trials peaked at the first recovery step while SW for the mixed perturbation trial peaked at the second.

While no clear anticipatory adjustments for the step of the perturbation were observed, participants in our study achieved a

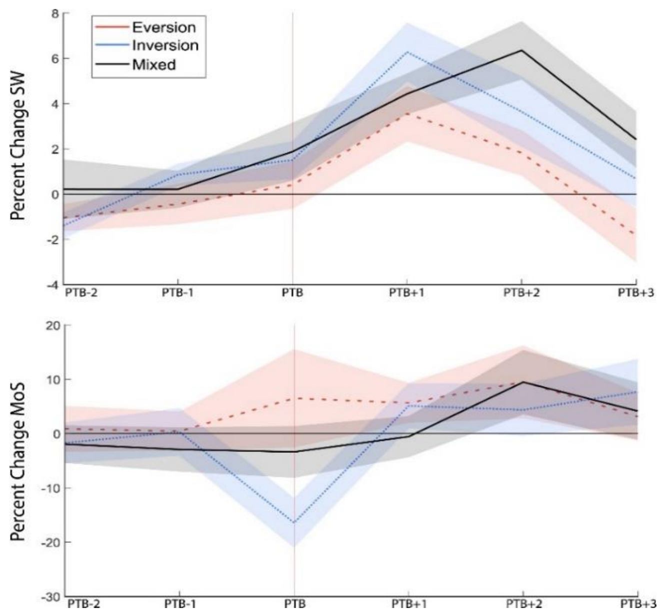

Figure 1: Mean (solid) and standard error (shaded) percent change to SW and MoS across anticipatory and recovery steps. Participants completed trials with eversion perturbations (red), inversion perturbations (blue) and randomly switching eversion and inversion perturbations (black).

normal SW sooner when the perturbation had predictable physical features, suggesting they were able to prime a response when the perturbations were predictable. Our results bear resemblance to studies of standing balance for which individuals can rely less on compensatory balance mechanisms when an upcoming perturbation has both predictable magnitude and timing[3]. A primary limitation of this study is the unilateral perturbation, which participants may have mitigated with other locomotor strategies such as reducing the load on the affected limb. Future work should investigate strategies during overground locomotion or walking with turns.

Significance

Healthy young adults recover normal gait following underfoot perturbations faster when perturbations have predictable physical qualities. Individuals with difficulties integrating proprioceptive information may struggle to recover balance in complex walking environments due to an inability to interpret differences between the predicted and experienced shifts to underfoot CoP.

Acknowledgments

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COMPARISON OF OBSTACLE CLEARANCE UNDER MIXED, VIRTUAL AND PHYSICAL REALITY ENVIRONMENTS

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Introduction

Fall-risk associated with tripping over obstacles is a serious concern among the several populations like older adults.[1] Hence it is critical to identify assessments and mechanisms that increase this fall-risk to design effective fall-prevention programs. Biomechanics of obstacle clearance is a well-studied topic as it gives in-sight into successful factors associated with avoiding tripping. Previously, obstacle clearance/avoidance was studied by using obstacles of different heights, surfaces and so on in physical reality (PR).[2] A relatively novel way of studying obstacle clearance is to use immersive technologies like Virtual Reality (VR) and Mixed Reality (MR) in which users wear a head mounted display and visualize and interact with computer-generated contents. While VR offers a totally immersive environment, MR offers a hybrid environment where participants can see the real/physical environment they are in while interacting with virtual objects.

The purpose of our study was to compare the obstacle clearance strategies used by healthy young adults in these three environments. We hypothesized that the obstacle clearance strategies of both leading and trailing legs in MR will be intermediate of the strategies used in VR and PR environments.

Methods

Twelve healthy, young participants walked and crossed over MR, VR, and PR obstacles set at 2” (5.1cm), 6” (15.2cm), and 15” (38.1cm). These functional heights were chosen to represent a door threshold, stairs, and the first step into a bus respectively. The order of obstacle heights and environments were randomized between the participants. Three trials per obstacle were performed. Reflective markers were placed bilaterally as follows: 2nd metatarsal head and posterior calcaneus. The level of obstacle clearance for each trial was quantified by the peak height of the toe and heel markers of both the legs. An average of the three trials was used for statistical analyses. A two-way within-subjects ANOVA with height (2” vs 6” vs 15”) and environment (MR vs. VR vs. PR) was performed.

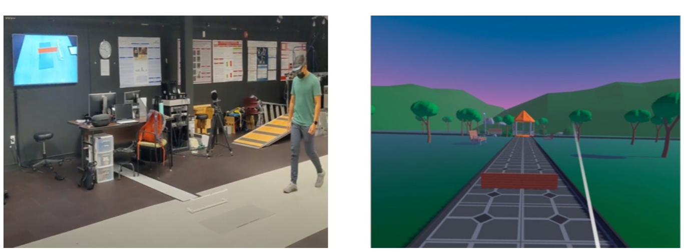

Fig.: Obstacle crossing in MR (left) and VR (right) environments

Results and Discussion

Significant interaction for heel clearance of lead ($P = 0.022$) and trail ($P < 0.001$) legs and toe clearance of trail leg ($P < 0.001$; Table 1). There was a significant main effect of environment for lead leg toe and heel height in MR was significantly small than that VR and PR. Significant main effect of height showed that as the height of the obstacle increased, participants had greater toe and heel clearance across all the environments.

Results were in contrast to our hypothesis where performance in MR environment was either too conservative (foot clearance was way higher than the obstacle) or too risky (foot clearance was too close to the obstacle) depending on the obstacle height. In general, while going over the 2”, 6” and 15” obstacles with their leading leg, participants overestimated height in the virtual environments (MR and VR) compared to the PR. Moreover, participants took a more conservative approach to raise their leg higher in MR environment compared to VR environment. This was largely true for the trailing leg as well but for the 15” obstacle where participants did not raise their trailing leg as high in MR and VR compared to PR environments. So, while clearing the highest obstacle, it seemed that the participants focused on only successfully clearing the obstacle with their leading leg but not with their trailing leg. This could be due to the potential novelty and level of immersion felt within virtual environments.

Table 1. Mean (SD) foot clearance values

Variable	OH	MR	VR	PR
LL PTH (cm)	2”	36.6 (6.8)	33.5 (5.3)	21.3 (2.0)
	6”	45.9 (8.9)	42.2 (6.9)	31.6 (3.1)
	15”	68.8 (8.1)	68.9 (11.3)	57.5 (4.2)
LL PHH (cm)	2”	50.7 (6.0)	46.5 (5.8)	34.9 (2.7)
	6”	54.2 (5.3)	51.9 (6.9)	43.6 (4.4)
	15”	67.0 (4.6)	66.4 (7.1)	60.4 (3.6)
TL PTH (cm)	2”	30.5 (11.7)	28.9 (9.8)	22.3 (3.1)
	6”	40.6 (14.4)	37.0 (12.0)	33.5 (6.5)
	15”	52.9 (15.7)	60.9 (16.1)	62.0 (4.5)
TL PHH (cm)	2”	49.3 (13.7)	49.4 (11.4)	43.1 (4.8)
	6”	59.0 (14.6)	58.0 (13.3)	56.0 (6.6)
	15”	70.8 (12.2)	78.9 (16.0)	83.2 (3.3)

OH – Obstacle height; LL – Lead Leg; TL – Trail Leg; PTH - Peak Toe Height; PHH - Peak Heel Height

Significance

Studies have shown beneficial applications of these virtual environments as effective interventions esp. when working with older adults.[3] While VR has been used more for training purposes and studies using MR are scarce, there has been limited use of both VR and MR as a tool for physical therapy (PT) assessment.[3] The ability to use a VR/MR apparatus could potentially be a more cost-effective tool and be able to clear up clinic space as well as be customized to fit different PT assessment needs. Using virtual obstacles in real environment can also minimize the risk for causing actual fall due to limited clutter during patient-therapist interactions. Results of the current study could help in creating such virtual environments.

References

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