Friday May 21st			
Start Time (UTC)	**Finish Time (UTC)**	**Event**	**Presenter**
5:00pm	**5:45pm**	**Poster session #3**	
	Walker	WALKING SPEED ALTERS BAREFOOT GAIT COORDINATION AND VARIABILITY	Binnan Yu
	Murray	ANKLE STIFFNESS MODULATION DURING DIFFERENT GAIT SPEEDS IN INDIVIDUALS POST-STROKE	Erica Hedrick
	Glassow	BRAIN ACTIVATION DURING SINGLE AND DUAL TASK WALKING IN PEOPLE WITH PARKINSON'S DISEASE	Becky Wagner
	Gilbreth	EFFECT OF ARM CONSTRAINT ON LEADING LIMB INTER-SEGMENT COORDINATION STABILITY DURING OBSTACLE CROSSING IN YOUNG ADULTS	Ho-Cheng Lu
	Walker	MOBILE, BIOSENSOR TECHNOLOGY FOR MEASURING JOINT-LEVEL HUMAN MOTION	Matt McManigal
	Lee	SEQUENTIAL AUTO-IMITATION LEARNING OUTPERFORMS FEEDBACK-ERROR LEARNING	Maximilian Stasica
	Murray	A SYSTEMATIC REVIEW: LONG RANGE CORRELATIONS IN RUNNING GAIT	Taylor Wilson
	Glassow	SUPERVISED EXERICSE LEADS TO MEANINGFUL IMPROVEMENTS TO WALKING SPEED IN PATIENTS WITH PERIPHERAL ARTERY DISEASE	Hafizur Rahman
	Lee	IMPLEMENTATION AND EVALUATION OF A BASEALL PITCHING PROGRAM AND ITS IMPACT ON INJURY PREVENTION AND PERFORMANCE	Tyler Hamer
	Murray	IMPROVING MOBILITY IN PERIPHERAL ARTERY DISEASE USING AN ANKLE FOOT ORTHOSIS: EFFECT OF PHYSICAL ACTIVITY ON THE METABOLIC COST IN PATIENTS WITH PERIPHERAL ARTERY DISEASE WHILE WALKING WITH AND WITHOUT AN ANKLE FOOT ORTHOSIS	Nate Evans
	Lee	A WII BALANCE BOARD CAN CAPTURE CHANGES IN POSTURAL DYNAMICS RESULTING FROM TASK MANIPULATIONS DURING STANDING	Takashi Sado
	Walker	SOMATOSENSORY-MOTOR INTERACTIONS DURING FINE AND GROSS GRASPING	Madison Davis

Poster rooms shown in the right column above next to the poster title.

WALKING SPEED ALTERS BAREFOOT GAIT COORDINATION AND VARIABILITY

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Presentation Preference: [**Poster** and **Podium**]

INTRODUCTION

Barefoot walking is advocated as an exercise due to its potential to improve foot muscle strength and decrease the load in the knee joint[1]. This study examines the inter-joint coordination and variability of barefoot walking. We hypothesize that 1) walking speeds influence the similarity of coordination pattern between barefoot and shod, and 2) coordination variability is lower in barefoot than shod walking at all walking speeds.

METHODS

20 healthy female subjects were recruited and performed walking trials in each speed category (SLOW, CWS, FAST) and both barefoot and shod walking. Three couples are selected: ankle-knee, knee-hip, and ankle-hip. Inter-joint coordination is obtained using the vector coding method[2]. Cross-correlation coefficients (CCC) and root-mean-square-differences (RMSD) are used to determine the similarity of coordination patterns between shod and barefoot walking in each of three speed categories. If a global effect of walking speed is detected in a specific couple, we further categorize it into 4 different patterns[3]. Coordination variability is estimated using angular deviation (AD). CWS is defined as comfortable walking speed.

RESULTS AND DISCUSSION

The ankle-knee pattern becomes more similar (CCC & RMSD: $p<0.00001$), and the ankle-hip coordination becomes less similar (CCC & RMSD: $p<0.001$) between shod and barefoot as walking speed increase (Figure 1). This result suggests that barefoot and shod walking use different coordination strategies to cope with the increased walking speed (Figure 1-2).

In barefoot, AD is significantly lower in the late swing of the ankle-hip couple at SLOW ($p < 0.01$) and CWS ($p<0.05$). Without the cushioning of athletic shoes, individuals may subconsciously adopt other strategies to cope with the impact. We postulate that the decreased coordination variability in the late swing may be the consequence of intentional muscular control to prepare for the anticipated impact at the heel strike.

AD is also significantly lower in the mid-stance of the knee-hip couple during CWS ($p<0.05$) and FAST ($p<0.05$). A reduced coordination variability, or a more repretitive motor pattern, can introduce higher stress into the joint tissue in the long term[4]. One possible explanation of the reduced coordination variability in barefoot is modern humans have adapted to the shod condition during evolutionary timescale, and walking barefoot is no longer "natural", as claimed by anthropologist[5].

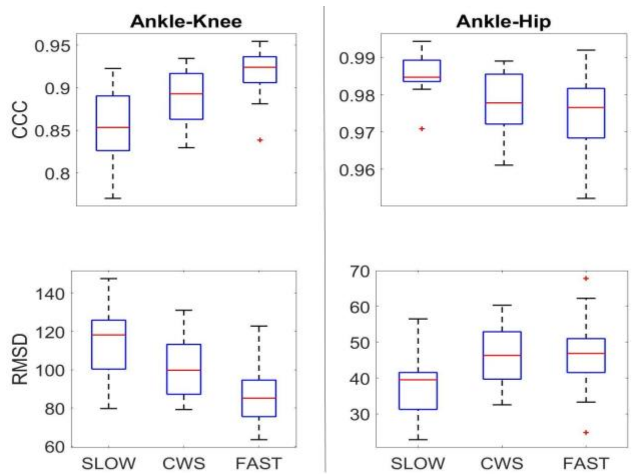

Figure 1: Similarity of coordination pattern at three speed-categories. CWS: comfortable walking speed.

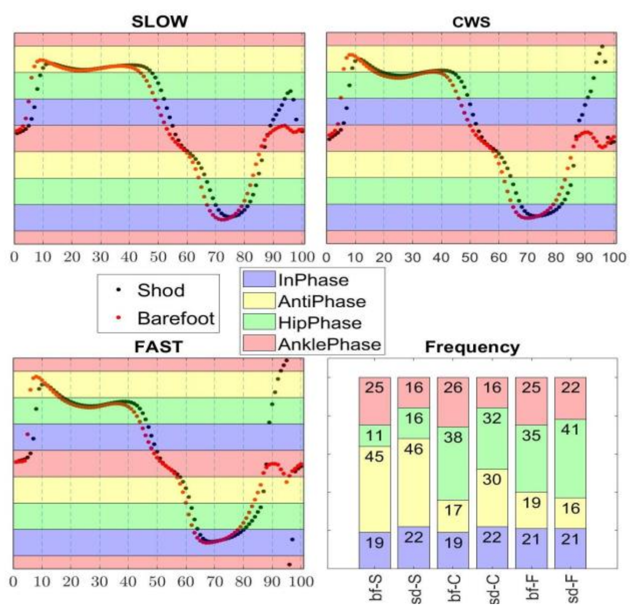

Figure 2: Ankle-Hip Pattern Category. bf: barefoot; sd: shod; S: SLOW; C: CWS; F: FAST. The horizontal axis for SLOW, CWS and FAST plots is stride cycle with the unit in percent.

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ANKLE STIFFNESS MODULATION DURING DIFFERENT GAIT SPEEDS IN INDIVIDUALS POST-STROKE

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Presentation Preference: **Poster**

INTRODUCTION
Ankle joint stiffness contributes to walking performance in individuals after a stroke and is related to maximum ankle joint power at push off on the paretic side [1]. Healthy individuals can alter their ankle joint stiffness in response to changing walking speed [2], however, for individuals post-stroke the ability to alter their ankle stiffness when increasing walking speed is unknown. Individuals post-stroke commonly have weak plantar flexors due to a decreased force generating ability. Plantar flexor weakness post-stroke may limit the ability to alter ankle stiffness as healthy individuals do, resulting in alternate compensation patterns. The purpose of this study was to investigate the relationship between ankle stiffness and walking speed and propulsion, at two walking speeds. We hypothesized that individuals post-stroke will not increase paretic ankle stiffness from self-selected to a fast speed due to a limited capacity to increase ankle moment. Additionally, we hypothesized that ankle stiffness will correlate with gait speed and propulsion within each speed condition.

METHODS
Twenty-nine participants (15M, 14F; age 58.3 ± 13.1 yrs; 4.8 ± 7.8 years since stroke (range=0.5 - 30.5 years); 15 R/14 L paresis) from a larger study conducted at the University of Delaware were analyzed. Motion capture was performed on an instrumented treadmill while participants walked at two speeds: their self-selected speed (SS) and a fast walking speed (FS). Paretic and nonparetic ankle joint 'quasi-stiffness' was calculated during the dorsiflexion stance phase as the slope of the ankle moment versus ankle angle curve [3]. Peak paretic and nonparetic propulsion were calculated using the peak anterior ground reaction force during the second half of stance. A 2x2 repeated measures ANOVA was performed for ankle stiffness with side (paretic or nonparetic) and speed (SS or FS). Additionally, a linear regression was performed between speed and paretic (or nonparetic) ankle stiffness and propulsion and paretic (or nonparetic) ankle stiffness for both the SS and FS.

RESULTS AND DISCUSSION
There was a main effect of side (F=5.515, p=0.026) but not speed (F=0.48, p=0.365) on ankle stiffness (Figure 1). The interaction of side and speed on ankle stiffness was not significant (F=0.011, p=0.919). Specifically, paretic ankle stiffness was less stiff than the nonparetic ankle stiffness for both the SS and FS. These results contradict a previous study that found no difference between paretic and nonparetic side [1]. Reasons for the conflicting findings could be the previous study's low sample size (N=12) compared to our study (N=29) or the inclusion of people who have survived either a brain tumor or a stroke while our study solely evaluated individuals after a stroke. There may also be differences in stroke severity between studies but were not reported in the previous study.

Ankle stiffness does not change between speed conditions. A previous study suggests that the healthy human ankle changes from a passive to an active system during increased walking speed [2]. Individuals post-stroke use more of the passive component of stiffness on their paretic side [4], which could suggest that it is difficult for them to change to an 'active system', causing them to not be able to increase their stiffness and may limit their walking speed.

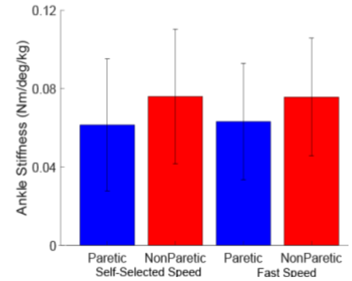

Figure 1: Ankle stiffness is greater on the nonparetic (red) compared to the paretic side (blue).

Although there were no increases in stiffness for either extremity across the speeds tested, within each condition there was a positive relationship between ankle stiffness and walking speed and ankle stiffness and peak propulsion across individuals (Table 1). This relationship between stiffness and self-selected speed has been reported previously [1].

Table 1: R squared values for the relationship between ankle stiffness (paretic or nonparetic) and speed or peak propulsion (AGRF) (*p<0.05, **p<0.01)

R^2 values	Self-Selected Speed		Fast Speed	
	Paretic	Non Paretic	Paretic	Non Paretic
Speed	0.28**	0.32**	0.23**	0.27**
Peak AGRF	0.25**	0.17*	0.23**	0.17*

CONCLUSIONS
Understanding post-stroke ankle stiffness may be important in the design of ankle-foot orthoses or exoskeletons, where these devices can augment the biological stiffness seen at the ankle to improve walking performance. The relationship between ankle stiffness and speed and peak propulsion suggests that increasing ankle stiffness could increase an individual post-stroke's walking speed or propulsion. Additionally, understanding an individual's ankle stiffness profile may aid with clinical rehabilitation decision-making for targeted interventions.

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BRAIN ACTIVATION DURING SINGLE AND DUAL TASK WALKING IN PEOPLE WITH PARKINSON'S DISEASE

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Presentation Preference: **Poster**

INTRODUCTION

Parkinson's Disease (PD) is a degenerative disease that effects the dopamine-producing neurons of the brain, resulting in tremors, limb rigidity, and gait and balance problems.[1] In the present study, gait and balance problems were evaluated under the strain of dual-task walking. The introduction of the second task moves key attentional resources away from walking, which can result in an increased risk of falls, especially for older adults with PD.[2] Previous studies have shown increased prefrontal brain activity in people with PD, even during simple walking[3], which suggests it is a good indicator of "attentional resources consumption". Prefrontal activity also increases during dual-task walking, however in these tasks it is unclear if the participants focus more on the locomotor or the cognitive task. In this study, we propose to compare "uninstructed" to "prioritized" dual-task walking. We have three hypotheses: 1) prefrontal activity will be higher in people with PD in all conditions; 2) instructing people with PD to focus on the cognitive task will increase prefrontal activity compared to when they are "uninstructed"; 3) instructing people with PD to focus on the walking task will show similar levels of prefrontal activity compared to when they are "uninstructed".

METHODS

While the proposed study will include 60 PD patients and 60 age-matched controls, the current analysis was over 8 participants (4 PD, 4 healthy controls). Participants completed a series of clinical tests prior to being fitted with pressure sensitive insoles, inertial measurement tools, headphones, and a functional near infrared spectroscope (fNIRS). Following their adornment of the clinical measurement tools, participants were asked to complete walking and cognitive listening single tasks.

During the follow-up appointment, participants were introduced to the dual-tasking procedures by having to walk and complete the cognitive listening task at the same time. The first dual-task trial was completed with no instructed prioritization to one task or the other. Prioritized instructions for diverting attentional resources to the walking task or cognitive listening task were then randomized following the baseline task. Prefrontal activity was monitored using the fNIRS. Data collected on the prefrontal activity was then block averaged into two-minute windows and the first two minutes and last two minutes were compared to determine the attentional demands of completing the single and dual tasks.

RESULTS AND DISCUSSION

Based on the current data, we are seeing trends that align with previous literature and others that are more surprising. Since PD patients must dedicate more attention to their walking, we expect to see more prefrontal activity during walking tasks compared to healthy older adults, which we are observing in the walking single-task. and the cognitive-emphasis dual-task. However, some of the other walking tasks, no-prioritization and walking-prioritization dual-tasks in particular, are showing less prefrontal activity than we anticipate for PD patients compared to controls (Figure 1). We predict that this could be due to a small sample size and possible outliers.

In future analyses, we plan to analyze hemispheric activation across the tasks, expecting to see higher levels of activation for the left prefrontal cortex compared to the right in dual tasks due to the linguistic nature of the cognitive task. We also plan to split the measurement time even further in two-minute windows for the entire task to observe how prefrontal activity changes as the tasks progress.

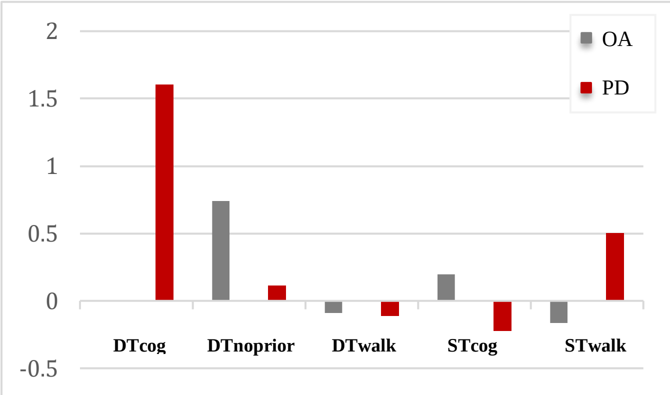

Figure 1

CONCLUSIONS

The activation of the prefrontal cortex increases with the amount of attention required to perform a task. In patients with PD, these resources are already dedicated to walking. By increasing the attentional load of a task, PD patients must pull attentional resources from their walking to the new task, putting them at a risk of falling.

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ACKNOWLEDGEMENTS

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EFFECT OF ARM CONSTRAINT ON LEADING LIMB INTER-SEGMENT COORDINATION STABILITY DURING OBSTACLE CROSSING IN YOUNG ADULTS

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INTRODUCTION

Adequate arm swing can contribute to whole body dynamic stability during obstacle crossing [1]. However, only a few studies have discussed the effect of arm constraint on lower extremity inter-segment coordination stability during obstacle crossing. Additionally, the previous studies demonstrated that the obstacle-crossing leading limb exhibited less inter-segment stability [2]. This reduced stability in the leading limb could indicate an increased difficulty in controlling the segment to modulate whole body stability. Hence, the purpose of this study was to investigate the effect of arm constraint on the leading limb's inter-segment stability during obstacle crossing. It was hypothesized that arm constraint would decrease the leading limb's inter-segment coordination stability by altering normal system dynamics and requiring novel compensatory behaviors.

METHODS

Eighteen subjects voluntarily participated in this study. Subjects walked 6 m at a self-selected speed in each of two arm constraint conditions, unconstrained and constrained, while stepping over a 20 cm height obstacle placed at the walkway's midpoint. Constrained arms were defined as the hands placed on a subject's waist without natural arm swing. Subjects were required to successfully perform five consecutive walking trials in each condition. Condition order was randomized. A customized lower body marker system adapted from the literature was used to determine retro-reflective marker placement for motion capture [3,4]. Raw data were recorded using an 8-camera Vicon motion analysis system (Nexus 2.9.3, Vicon Motion System, Centennial, CO; 100 Hz). Three-dimensional raw marker trajectories were low-pass filtered (6 Hz; Butterworth, 4th order) and then processed using a custom Matlab (The Mathworks, Inc., Natick, MA, USA) program that modeled the body and calculated sagittal plane segment angle and angular velocity variables.

The continuous relative phase (CRP) was calculated and used to represent inter-segmental coordination between adjacent lower extremity segments (i.e., pelvis-thigh, thigh-leg, leg-foot) in the sagittal plane. The CRP's variability, the deviation phase (DP), was used to quantify the inter-segment coordination stability [5]. A higher DP value indicates less stability in the movement [5]. After confirming parametric assumptions, a paired-samples t-test was used to compare leading limb DP values between unconstrained and constrained arm swing conditions during obstacle crossing. Statistical analysis was performed using SPSS (version 27.0, Chicago, Illinois).

RESULTS AND DISCUSSION

Descriptive statistics were calculated for the leading limb's DP variables in the two arm constraint conditions during obstacle crossing (**Table 1**). There were no significant ($p > 0.05$) arm constraint effects on the leading limb's inter-segment coordination stability when crossing the obstacle.

The results did not support the hypothesis that arm constraint would decrease the leading limb's inter-segment coordination stability when crossing the obstacle. Nakakubo [1] found similar results for the trunk with restricted arm swing during normal walking, but increased trunk stability with exaggerated arm swing. Bruijn [6] suggested that arm constraint might limit the recovery of overall stability during a gait perturbation. However, the current results suggest that any such constraint does not appear to specifically affect the leading obstacle-stepping limb's inter-segment coordination stability. It could be that the specific task was not demanding enough to elicit the hypothesized effect, or the effect could have occurred elsewhere in the body, such as the trunk, or in a non-sagittal plane. Additional work is necessary to investigate alternate hypotheses.

CONCLUSIONS

Constrained arm swing did not alter the leading limb's inter-segment coordination stability in the sagittal plane during obstacle stepping, and so leading limb coordination may not be a suitable intervention target to reduce coordination control-related fall risk.

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Table 1: Descriptive statistics for leading lower extremity (mean ± SD).

Swing Condition	Leading Limb Inter-Segment Coordination		
	Pelvis-Thigh	**Thigh-Leg**	**Leg-Foot**
Unconstrained Arms	25.3° ± 5.10	19.9° ± 3.55	12.5° ± 3.17
Constrained Arms	23.8° ± 4.62	19.5° ± 3.50	12.7° ± 3.43

MOBILE, BIOSENSOR TECHNOLOGY FOR MEASURING JOINT-LEVEL HUMAN MOTION

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Presentation Preference: **Podium**

INTRODUCTION

Real-time tracking of knee joint angle is of significant interest and can be used to predict ACL injuries and risk for developing early knee osteoarthritis [1,2]. However, assessment of knee joint angles is currently limited due to the lack of validated mobile technologies that can be deployed in rural or community settings. Here, we created a wearable electronic device to address this challenge and enable measurement of knee joint angles during natural human movement outside of a laboratory environment.

METHODS

The wearable device for measuring knee joint angles is composed of two inertial measurement units (IMUs, ICM20948, InvenSense), microcontroller (ATMega328, Microchip), 2.4GHz wireless transceiver (nRF24L01, Nordic Semiconductor), and rechargeable Li-Ion battery (3.7v, 500mAh) with power regulation (Figure 1). Each IMU contains a three-axis accelerometer, gyroscope, and magnetometer to track human motion. The final, miniaturized device is shown in Figure 1B and will be integrated with a kinesiology tape substrate using a thermoplastic polyurethane heat sensitive film to allow the device to be temporarily attached to the leg for hours to days.

For data collection, the IMU sensors are arbitrarily placed above and below the knee joint. A calibration procedure is then performed through flexion/extension of the knee and abduction/adduction of the hip [3]. A quaternion-based, strap-down gyroscope integration algorithm is used to estimate joint angle [4]. The accelerometer data is used to reduce gyroscope drift during low frequency movements. During operation, motion data from the gyroscope and accelerometer is collected from each IMU at a frequency of 150Hz, these data are then wirelessly transmitted to a local computer, and knee joint angle is estimated in near real-time. The wearable device is capable of continuously collecting motion data from the IMU sensors for up to 2 hours.

RESULTS AND DISCUSSION

Preliminary testing indicates that our device is able to measure sagittal plane knee angles within known anatomical limits of knee flexion and extension (Figure 2).

CONCLUSIONS

We have created a wearable device that can wirelessly collect human motion data and estimate knee joint-angle in near real-time outside of a laboratory environment. Future work will include optimization of the wearable device and validation of the knee joint angle algorithm against a 3D-motion capture system, the current gold-standard for measuring knee joint angle.

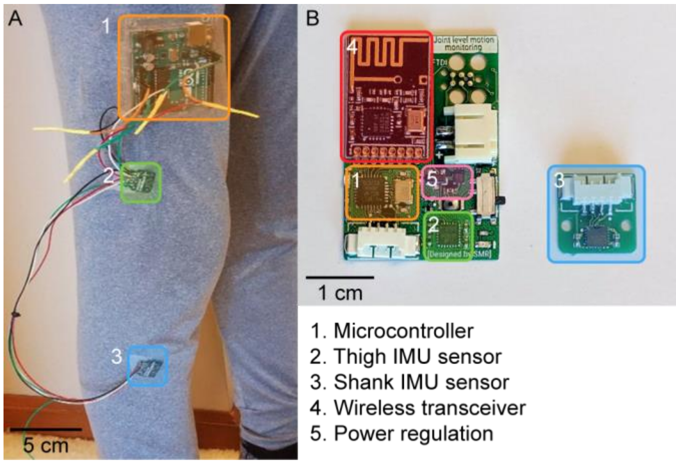

Figure 1: (A) Photograph of the sensor placement above and below the knee. (B) Miniaturized, printed circuit boards with microcontroller, IMU sensors, wireless transceiver, and power regulation.

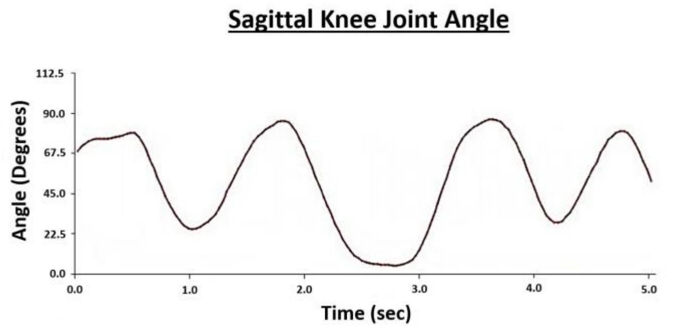

Figure 2: Estimation of sagittal knee joint-angle data during arbitrary knee flexion/extension between 0-90° flexion.

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SEQUENTIAL AUTO-IMITATION LEARNING OUTPERFORMS FEEDBACK-ERROR LEARNING

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Presentation Preference: [**Poster, Please consider for Promising Graduate Student Award**]

INTRODUCTION

Contrary to robots, humans require an inverse dynamics model to control their limbs [1]. Thus, the question arises, how such a model can be acquired. Two of the currently discussed models, which enable learning of such an inverse model, are Feedback-Error Learning (FEL) and Auto-Imitation Learning (AIL). FEL uses the tracking error to learn the inverse dynamics [2] and therefore can be seen as a form of goal-directed learning. AIL is essentially a parallel system-identification process [3] and can be filed as non-goal-directed learning.

METHODS

In the context of the test-trilogy [4], a simulation test checks whether a concept is mathematically sound and performs as expected. We therefore conduct a simulation study using a two-joint controller-plant system with an upper and a lower joint.

Based on redefined Hebbian Learning [5] we derive a sequential AIL-architecture (sAIL) to compare it under the same conditions to the also sequential FEL. The sAIL contains an operator and a learner with the same model structure, respectively, which enables forwarding the model weights to the operator (Fig. 1).

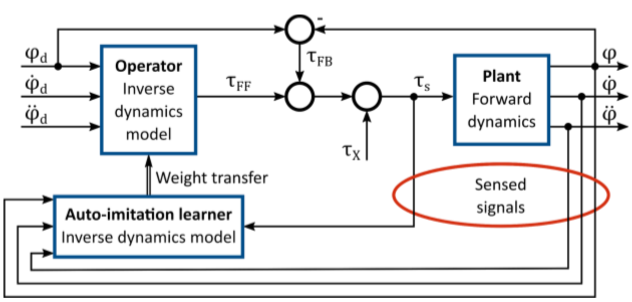

Figure 1: The operator uses its inverse model of the plant to determine the required feedforward torque τ_{FF} from the desired angular kinematics (φ_d, $\dot{\varphi}_d$, $\ddot{\varphi}_d$). The actually sensed signals τ_s, φ, $\dot{\varphi}$, and $\ddot{\varphi}$ at the plant flow into the learner which transfers the updated weights back to the operator.

The plant comprises a peripheral feedback delay of $\Delta t = 0.01$ s and a learning rate of $r = 10$. The learning gains of the feedback errors are set to $K_{p1} = K_{p2} = 3$ for position and $K_{v1} = K_{v2} = 2$ for velocity. The simulation lasts 80 seconds. The first intervention is set at 12 s (a perturbation $\tau_X = 2$ Nm for 1 s on the upper joint). The second intervention is set at 22 seconds (a change in viscous damping of the upper joint by -1 Nms/rad).

RESULTS AND DISCUSSION

During the first twelve seconds both algorithms produce perfect results and do not change their weights. After the first intervention, sAIL does not change the weights, while FEL does, even though a short perturbation should not result in a weight change. After the second intervention sAIL reduces the respective weight to regain stability, while FEL becomes unstable.

In the FEL-scenario the weights of the lower joint change although the interventions are set to only target the upper joint. In contrast, sAIL keeps the weights of the lower joint constant as we expected.

CONCLUSIONS

Contrary to [6, 7], we consider the difference in delay between target and feedback paths. Compared to the target path, the feedback path introduces additional delays causing these two to get out of sync. Thus, in FEL, the feedback error and the re-adaptation are distorted while sAIL is capable of producing stable results nonetheless. As biological systems are always subject to such delays, the simulated FEL-architecture is not able to explain human motor control.

We conclude that out of the two simulated models, only sAIL can be considered a possible explanation for human motor control. Further investigation in the sense of the test-trilogy [4] may confirm the functional validity of this approach in form of a hardware test followed by a behavioral test on the human.

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A SYSTEMATIC REVIEW: LONG RANGE CORRELATIONS IN RUNNING GAIT

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Presentation Preference: [**Podium**]

INTRODUCTION

Much research has been dedicated to measuring the health benefits of running, but only a few studies have examined how running patterns change over time. Long range correlations (LRCs) characterize the degree to which movements are correlated from one moment to the next. LRCs are found in many human activities, including walking and running. Assessing LRCs is important because the presence of LRCs has been associated with health, while the absence of LRCs has been noted as a marker of disease [2]. Hence, more research is needed to understand the meaning of LRCs in running gait.

A systematic review (SR) was completed to look at the effects of running gait on LRCs and the implications this could have on human health, performance, and rehabilitation. The aims of this review were to: identify the typical LRC patterns in human running gait, the effect that running/walking has on LRCs, the effect of injury/disease on running gait, and the effect that surface has on the LRCs of running gait.

METHODS

PRISMA guidelines were followed for this SR. PubMed, IEEExplore, Scopus and Web of Science were used until November 2020 using the SR tool Rayyan Qatar Computing Research Institute.

Inclusion criteria were: Experimental studies involving human subjects, running, and measurements of LRCs. Exclusion criteria were: Non-experimental studies, non-humans, walking only, non-running, non-LRC analysis, and non-experiments. Two independent raters (TW,AL) screened articles for inclusion. Raters settled disagreements through discussion.

RESULTS AND DISCUSSION

200 studies fell within our criteria. After review and deliberation between TW and AL, 20 articles were included in our SR. Our review revealed that, in general, gait characteristics of running seem indicative of LRCs and are similar to those found in walking gait. LRCs are apparent in both treadmill (TR) and overground running (OR). Results comparing OR and TR, however, are mixed, with some papers showing TR produces greater LRC [5], while others show the opposite trend [3].

Surprisingly, we did not find any studies comparing LRC as a function of health; however, two articles directly measured the effect of injury on the LRCs in running gait. One of those articles found only small differences in LRCs between injured and non-injured runners [6], compared to a significant difference in LRCs between the two groups, with a higher LRC in the non-injured runners [7]. Interestingly, the fatigue state of the runner may determine the magnitude of LRCs: One article showed a U-shaped trend in LRCs with highest levels of LRCs at the beginning and end of a run [8]. Other articles showed a linear decrease in LRCs with fatigue [1]. Running speed may produce a similar U-shaped LRC trend in running gait [5], but this effect was not consistent in this review [6].

CONCLUSIONS

We conducted this review to uncover the typical LRCs of running gait in several contexts. With mixed results, our review suggests that, like walking, running gait exhibits patterns of LRCs that seem to vary according to where running takes place and the state of the runner. If, as our review implies, fatigue produces measurable changes in LRCs, then tracking LRCs in a workout program could enhance injury prevention and speed up recovery. For instance, decreases in an athlete's LRC during training could be used to trigger suggestions for more rest and, ultimately, prevent injury. Our results also suggest new training possibilities that could enhance adaptability of the system. Training outside the preferred speed may increase LRCs, which could, in turn, improve recovery times and prevent injury. Furthermore, measuring LRCs of locomotion is highly recommended due to its good reliability [4], and non-invasive nature compared to other methods of recording fatigue, like taking lactate levels and measuring intramuscular EMG signals during a run.

More research is needed to determine how LRCs depend on running surface. This could have implications in sports performance and rehabilitation settings. By determining the surface that produces the highest LRCs, this will benefit coaches/physical therapists so that they can prescribe appropriate surfaces for their athletes/patients to improve healthier functioning. Lastly, our review suggests a strong need for further research into the timing of gait when running, due to mixed results concerning running surface, no research related to disease and running, and the relatively low number of papers that have investigated LRCs in this context.

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SUPERVISED EXERICSE LEADS TO MEANINGFUL IMPROVEMENTS TO WALKING SPEED IN PATIENTS WITH PERIPHERAL ARTERY DISEASE

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Presentation Preference: **Poster**

INTRODUCTION

Peripheral artery disease (PAD) is a manifestation of systemic atherosclerosis affecting the leg arteries and causes walking induced pain known as intermittent claudication. Supervised exercise therapy (SET) is a first-line treatment for patients with PAD leading to significant improvement in the distances patients can walk and in quality of life [1]. The efficacy of SET is most commonly expressed by significant statistical improvement of parameters from quality of life questionnaires, walking distances, and gait biomechanics. However, these comparisons do not quantify how each individual patient will benefit from SET, or the practical significance of improvements. The minimal clinically import difference (MCID) represents the smallest change in an outcome measure that is significant and relevant to patients [2]. This study estimated the MCID in walking speed in patients with PAD after SET.

METHODS

Patients and Experimental Data Collection

A total of 63 patients (age: 64.95 ± 6.60 years) with Fontaine stage II PAD were recruited through the Nebraska Western Iowa Veteran Affairs Medical Center and University of Nebraska Medical Center. Each patient participated in a six-month SET program based on the American College of Sports Medicine Recommendations and were evaluated: 1) before (*baseline*) and 2) after six-months (*post-exercise*) of SET [1]. During experimental testing, patients walked across a 10-meter pathway while a reflective marker was placed on the heel of the leg most affected with the PAD. The coordinates of the marker were tracked using a 12-high speed infrared camera system (60 Hz, Motion Analysis Corporation, CA). Walking speed was calculated as the average distance traveled per second as measured by the reflective marker.

MCID Calculation

For the distribution-based method, small and substantial improvements were computed as 0.2 and 0.5 x σ, respectively; σ is the standard deviation of the baseline walking speed [2]. For the anchor-based method, we used the mobility question “ability to walk one block” from the SF-36 questionnaire. Patients were categorized into substantial, small, or no improvement based on their rated ability as *limited a lot*, *limited a little*, and *not limited at all* while answering the anchor question [2]. Mean change in walking speed between these groups was reported as an estimate for the MCID in walking speed. A receiver operating characteristics (ROC) curve was used to estimate the threshold walking speed to predict improvement after SET (any improvement versus no change).

RESULTS AND DISCUSSION

The average walking speed of all patients at baseline was 1.11 ± 0.15 m/s. Walking speed increased to an average of 1.16 ± 0.16 m/s after six-months of SET (4.5% improvement, $p < 0.001$). Based on standard deviation of walking speed at baseline (0.15), the distribution-based method estimated a change of 0.03 m/s for small improvement and 0.08 m/s for substantial improvement after SET. There were 38.1% of patients with a small improvement and 6.3% had a substantial improvement in walking one block. Small and substantial improvements according to the anchor question “walking one block” were 0.05 m/s and 0.15 m/s, respectively. ROC curve analysis identified an increase of 0.04 m/s for improvement based on walking one block (Fig. 1)

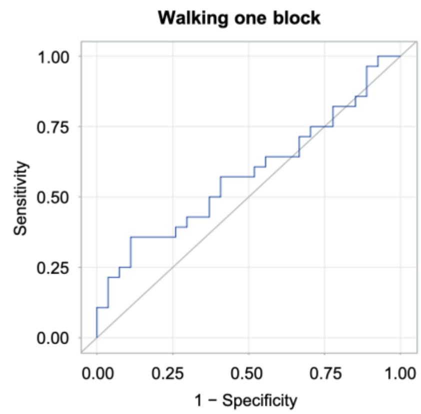

Figure 1: Threshold change in walking speed was 0.04 m/s (sensitivity 57.1%, specificity 44.4%, positive likelihood ratio 1.03, and area under the ROC curve 0.58).

The MCID reported in this study can serve as a benchmark for clinicians to develop treatment goals and interpret clinically meaningful progress in the care of claudicating patients with PAD. Future studies including greater number patients and selecting additional anchor questions may further improve the process of estimating the MCID for patients with PAD.

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IMPLEMENTATION AND EVALUATION OF A BASEALL PITCHING PROGRAM AND ITS IMPACT ON INJURY PREVENTION AND PERFORMANCE

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Presentation Preference: **Poster**

INTRODUCTION

An average of 25,000 baseball players compete in the National Collegiate Athletic Association (NCAA) every year [1]. Furthermore, 40% of NCAA baseball players will be injured at some point during the season [2]. Key indicators within the pitching motion allow us to monitor player mechanics and improve deficiencies when they are present. Quantitative data from these indicators help monitor performance to screen for signs of kinetic and kinematic deficiencies. The goal of this study was to examine the efficacy of biomechanical evaluations on collegiate baseball pitchers. Through this, we seek to develop a model for identifying at-risk athletes through a longitudinal assessment of pitching mechanics spanning pre-season to post-season along with in-season tracking of pathomechanics.

METHODS

Pitchers from a local NCAA men's baseball team were recruited to attend a pre- and post-assessment session spaced three months apart. Athletes were outfitted with a full-body retroreflective marker set before the acclimation and warm-up session. Once ready, the pitcher was instructed to throw from an indoor, force-plate instrumented mound towards a target located 56' away. Professional grade motion-capture cameras (Qualisys, Gothenburg, Sweden) were mounted on tripods placed symmetrically around the pitching mound to capture athlete motion. Camera sampling frequency was set to 240 Hz. Key biomechanics variables were calculated using Visual 3D (C-Motion Inc., Rockville, MD) software.

Upon completion, members of the research team identified key biomechanical variables previously shown to have the greatest impact on a pitcher's ability to transfer energy throughout the kinematic chain. A member of the research team met with each athlete to discuss where their pitching biomechanics were compared to a series of normative values for college pitchers. Specific pitching drills aimed at addressing each athletes' mechanical needs were prescribed for weekly completion. Recommended variables were marked as corrected if athletes improved pitching their biomechanics towards the normative range.

RESULTS AND DISCUSSION

Seventeen athletes were recruited between pre- and post-assessments. Seven athletes were excluded from the study due to missing either session. Recommended drills included Lead Leg Internal Rotation, Arm Patterning Progressions, Roll-In Progressions, and Rocker Drills. A total of 29 mechanical improvements were recommended, with all but one athlete receiving three to improve upon. Mechanical adjustments included *Shoulder Abduction, Pelvic Rotation, Pelvic Tilt, Trunk Rotation,* and *Stride*.

Athletes were able to correct 20 of 29 (69%) of recommended biomechanical adjustments. Pelvic Rotation, Trunk Rotation, and Stride exhibited the most improvement, with Shoulder Abduction exhibiting the least amount of improvement (Figure 1).

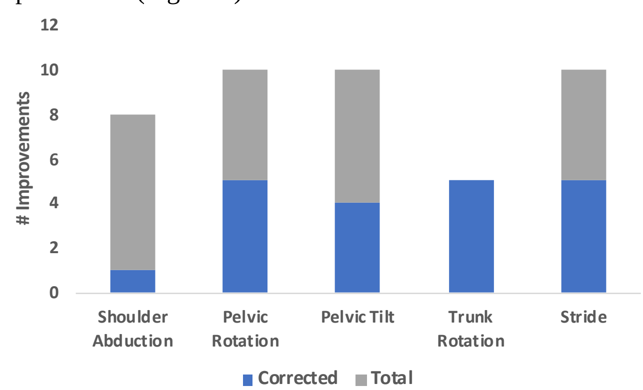

Figure 1: Post-assessment biomechanical improvements made compared to total pre-assessment recommendations

Results from this study showed that pitching biomechanics closer to the start of the pitching motion, such as Stride, Pelvic Tilt, Pelvic Rotation, and Trunk Rotation, had a higher frequency of being corrected compared to biomechanics closer to ball release (Shoulder Abduction). We believe this is due to the amount of time spent performing each action, where a pitcher spends roughly 0.5s in the early cocking phase, 0.11s in the late cocking phase, and 0.03s in the acceleration phase [3]. Previous research has also noted that the rapid amount of movement the arm makes over a short period of time therefore may make correcting upper extremity biomechanics more difficult [4].

CONCLUSIONS

Our comprehensive pitching evaluation demonstrates that 69% of biomechanical adjustments were able to be corrected within a three-month span. Pitching biomechanics closer to the start of the kinematic chain were found to be improved more than biomechanics towards the end. Further pitching evaluations will be conducted to observe the longitudinal impact on injury prevention.

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Improving Mobility in Peripheral Artery Disease Using an Ankle Foot Orthosis: Effect of physical activity on the metabolic cost in patients with peripheral artery disease while walking with and without an Ankle Foot Orthosis

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INTRODUCTION

Peripheral artery disease (PAD) is characterized as the buildup of atherosclerotic plaques preventing adequate blood circulation. PAD is most common in the lower extremities and it affects 20% of Americans 60 years and older[1]. The narrowing of blood vessels diminishes the oxygen delivered to the active muscles, which leads to claudication. An ankle foot orthosis (AFO) potentially reduces oxygen deficiency by substituting for the muscle force using the energy stored in the device. Overall metabolic oxygen consumption is a good indicator of how changes in biological muscle force impact overall energy cost of walking.

Previous studies have highlighted a consistent decrease in oxygen consumption by those wearing an AFO compared to those not wearing one[2]. The magnitude metabolic change with and without an AFO may be related to an individual's physical activity (PA). High levels of PA have been associated with increased efficiency of oxygen transport throughout the body, which results in an improvement in oxygen extraction capability[3]. We propose to investigate this relationship between PA and improvement in metabolic oxygen consumption during walking.

METHODS

Experimental data collection

A total of 25 subjects with PAD will be recruited from the claudication clinic at the Nebraska and Western Iowa Veterans Affairs' Medical Center in Omaha, NE. All subjects demonstrate positive history of chronic and exercise-limiting claudication and ankle/brachial index < 0.90 at rest.

Walking performance with and without wearing a carbon composite AFO will be measured. Subjects wear the AFO on both legs during AFO-trials. Measures of oxygen consumption will be taken during progressive treadmill trials for AFO and non-AFO conditions. During this portion of gait tests steady-state oxygen consumption is measured using a metabolic cart. Oxygen consumption during each stage of the treadmill test will be recorded as outcomes.

Physical activity will be measured by an Actigraph accelerometer that is worn on the subject's hip for seven days prior to the data collection session. From the accelerometer, average steps per day, maximum cadence, and average peak activity index will be calculated.

Data processing

Custom Matlab scripts will be used to calculate outcome measures from the oxygen consumption and PA data. Means and standard deviations will be calculated for oxygen consumption, walking distances, and physical activity outcomes.

Subjects will be divided into active and inactive groups based on their average steps per day. Average steps per day of 3586 will be used to make the distinction between low and high activity groups. Oxygen consumption and walking distance between the two groups will be compared between the AFO and non-AFO condition using independent t-tests. The change in oxygen consumption and walking distances between the two conditions will also be calculated and compared using paired t-tests.

EXPECTED RESULTS

The outcomes of this study will highlight a relationship between physical activity levels and the improvements in oxygen consumption during walking in patients with PAD. We will also learn how that relationship changes while wearing an AFO.

A significant positive correlation has been illustrated between average weekly sedentary time and maximum oxygen consumption in healthy, working adults aged 30-60[4]. Also, a decline in maximal delivery of oxygen is seen as a consequence of sedentary behavior in elderly groups[5]. For our study, we hypothesize that those with low levels of physical activity (average steps per day is <3586) or with lower walking performance/gait function will benefit more from using an assistive device such as an AFO and exhibit, on average, a lower rate of maximum oxygen consumption.

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A Wii Balance Board can capture changes in postural dynamics resulting from task manipulations during standing

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Presentation Preference: [**Poster**]

INTRODUCTION

Variability during postural sway has been shown to characterize healthy and pathological systems. Specifically, persistence from center of pressure displacement (i.e., how variability of postural control fluctuates over time) is a useful description of postural control in healthy populations [1] and in the investigation of pathologies such as Parkinson's disease and developmental disorders. However, obtaining such measurements outside of a laboratory setting is not simple. This is because portable laboratory-grade force platforms are expensive.

Our previous study showed that the Nintendo Wii Balance Board (WBB), a cost-effective portable device was able to accurately quantify persistence and anti-persistence behaviors of postural sway in the anterior-posterior direction compared to force platform (FP) in healthy populations [2]. However, typically studies use the center of pressure displacement. In a recent study, the point-to-point velocity of the center of pressure (COPv) was shown to be more sensitive in capturing the persistence and anti-persistence behavior in postural control [3]. Therefore, the purpose of this study is to determine the effects of task conditions on postural dynamics measured using a WBB.

METHODS

Seventeen healthy volunteers were recruited (3 females; mean age 24.5 ± 4.5 years). All participants performed three different conditions for 3 minutes: quiet standing on both legs with eyes open, quiet standing on both legs with eyes closed and standing on one leg with eyes open.

The WBB was placed centrally upon a flush mounted laboratory grade FP. The data was collected at 600 Hz for the FP, while it was collected at 30 Hz for the WBB. The data from FP was down-sampled to 30 Hz. From the COP displacement data, COPv was calculated for the WBB and FP. DFA was used to analyze the persistence and anti-persistence characteristics of the COPv in the anterior-posterior and mediolateral directions. First, inflection point was determined manually as the first time the persistence behavior shifted to anti-persistence behavior. Based on this, the alpha value of the short-term scale region (persistence) and long-term scale region (anti-persistence) was calculated. The level of statistical significance was set at $p < 0.05$.

RESULTS AND DISCUSSION

One-way repeated ANOVA revealed that there was a significant condition effect in alpha value of short scale region (alpha-short) in the AP ($p = 0.014$) and the ML ($p = 0.048$) directions. Inflection point, where the persistence behavior switched to anti-persistence behavior for the first time, also showed significant condition effect in AP ($p = 0.001$) and ML ($p = 0.014$). Post-hoc pairwise comparisons indicated that alpha-short for eyes closed standing was larger than one leg standing in AP ($p = 0.015$). Although there was no statistical significance ($p = 0.051$), it is interesting to see that in comparison to quiet standing, eyes closed standing has higher persistence. The eyes closed standing also showed earlier inflection point compared to the quiet eyes open standing ($p = 0.027$). In ML, one leg standing showed smaller alpha-short value compared to the eyes open quiet standing ($p = 0.004$), and had much earlier infection point compared to the eyes closed standing ($p = 0.044$). These results showed that WBB can capture the changes in persistence behavior of COPv between the standing conditions.

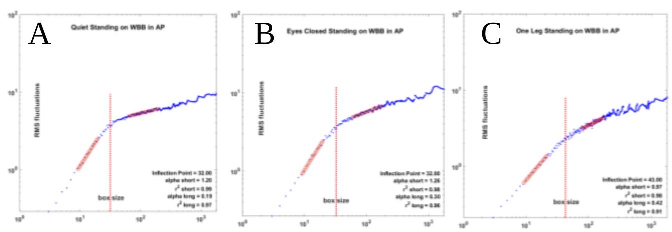

Figure 1A-C: Representative DFA data of COPv on WBB in AP of the same participant performing quiet normal standing with eyes open (1A), eyes closed standing (1B), and one leg standing (1C).

CONCLUSIONS

In this study, postural sway variability of COPv from WBB was analyzed using DFA. It was found that WBB can capture the changes of persistence in velocity of postural sway and is sensitive to task conditions. Future studies will compare how accurately WBB captures the persistence and anti-persistence of postural sway (COPv) compared to the gold standard, the FP.

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SOMATOSENSORY-MOTOR INTERACTIONS DURING FINE AND GROSS GRASPING

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<u>Presentation Preference: **Poster**</u>

INTRODUCTION

Our recent work has shown that afferent inputs to the primary somatosensory cortex (S1) are differentially processed during fine and gross grasping in humans (Lei et al. 2018). However, it remains largely unknown how S1 interacts with the primary motor cortex (M1) during these two grasping behaviors.

METHODS

<u>*Subjects*</u>. Seventeen healthy subjects participated in the study.

<u>*Experiment 1.*</u> Using electroencephalographic (EEG) and electrical stimulation, we measured somatosensory evoked potentials (SSEPs) component reflecting activation in the S1 during precision and power grip.

<u>*Experiment 2.*</u> Using transcranial magnetic stimulation (TMS) and electrical stimulation, we measured short-latency afferent inhibition (SAI) reflecting S1-M1 interactions via thalamo-cortical connections during precision and power grip. The TMS coil over the hand representation of M1 was oriented in the posterior-anterior (PA) and anterior-posterior (AP) direction to activate distinct sets of corticospinal neurons.

<u>*Experiment 3.*</u> Using dual-site TMS with specialized small coils, we measured intra-hemispheric interactions between S1 and M1 via cortico-cortical connections during precision and power grip.

RESULTS AND DISCUSSION

<u>*Experiment 1.*</u> The amplitude of the N20/P25 SSEP component was reduced during power compared with precision grip.

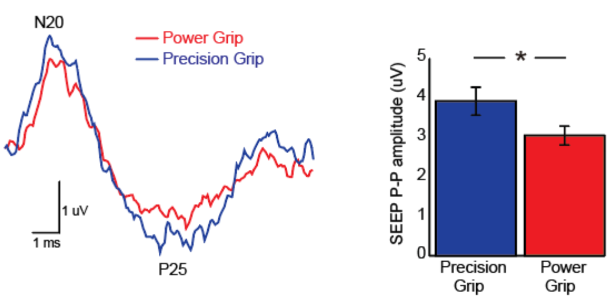
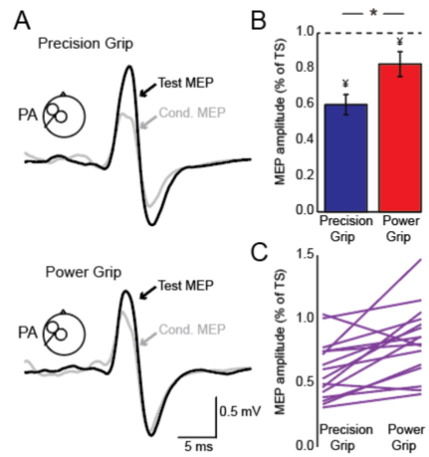


Figure 1: Raw SSEP traces recorded from the S1 in a representative subject during precision (blue) and power (red) grip (left). Group data show the amplitude of N20/P25 SSEP component (right).

<u>*Experiment 2.*</u> Afferent inputs attenuated motor evoked potentials (MEPs) to a larger extent during precision compared with power grip in the PA, but not AP, orientation.

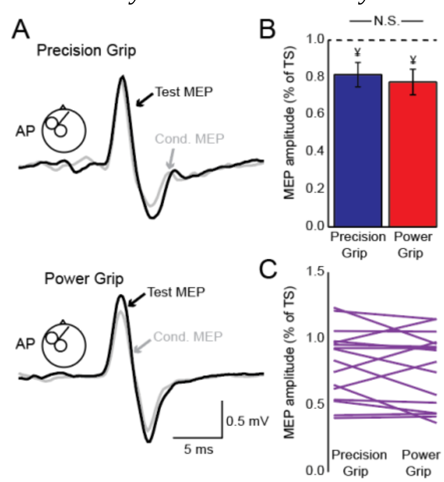
REFERENCE: Lei et al. J Neurosci 33, 7237-7247, 2018.

TMS-induced electric currents flowing from PA preferentially evoke highly synchronized corticospinal activity.



Figures 2: **A**. SAI was tested with the TMS coil in the PA orientation. **B**. Group data. **C**. Data from individual subjects.

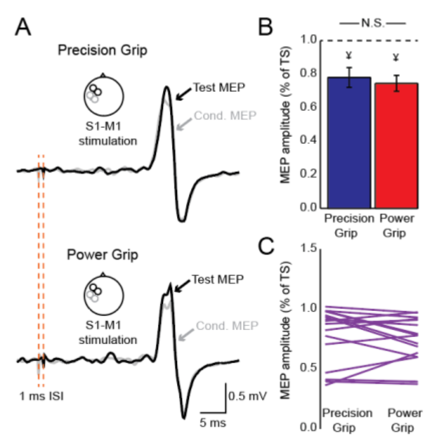
TMS-induced electric currents flowing from AP preferentially evoke less synchronized and delayed corticospinal activity.



Figures 3: **A**. SAI was tested with the TMS coil in the PA orientation. **B**. Group data. **C**. Data from individual subjects.

Figures 4: **A**. S1-M1 stimulation. **B**. Group data. **C**. Data from individual subjects.

<u>*Experiment 3.*</u> S1 stimulation reduced MEP during precision and power grip, while the ratio of S1-M1 inhibition remained similar between tasks.



CONCLUSIONS

Distinct S1-M1 circuits, probably from thalamo-cortical and cortico-cortical connections, contribute to fine and gross grasping in humans.