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Biomechanics and Neural Control of Movement: CMI's Effects on Downstream Motor Processing and Gait in Forwards and Backwards Walking

A Thesis Presented

by

Christopher Choi

To the Keck Science Department

Of Claremont McKenna, Pitzer, and Scripps Colleges

In partial fulfillment of

The degree of Bachelor of Arts

Senior Thesis in Neuroscience

April 29, 2020

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Abstract

Analyzing the effects of cognitive motor interferences (CMI) on walking is usually done in patients with neurological comorbidity or during forward walking (FW). However, there are few studies that examine gait differences between FW and backward walking (BW) under the presence of CMI when speed is kept constant on a treadmill. In this study we examined how CMI would disrupt sensory feedback and affect the descending motor pathway. We hypothesized that subjects that walked backwards and were given a cognitive task would show the greatest differences in gait due to a lack of visual input and the presence of CMI. A three-dimensional motion capture system was used to acquire the movement of the leg and calculate gait characteristics (stride length, stance phase, swing phase). Across the entire population, direction had a significant effect on all gait characteristics, but the presence of CMI did not have a significant effect on any of them. Additionally, there was no significant interaction between the two variables. Specifically, the overall stride was shorter, stance was shorter and swing was longer during backward conditions. However, within subject variability demonstrates that each subject utilizes different strategies to compensate for both the lack of sensory feedback and presence of CMI. Results of this study contradict findings from previous work that direction had no effect on stance and swing phase of walking and suggests that backward walking does change more gait characteristics. This implies that sensory feedback has a large impact on modulating motor output, and these effects may be amplified in those with movement-based neurological disorders like Parkinson’s Disease.

Introduction

Why Should We Study Movement?

For many, walking is a regular and unconscious part of life. People use it as a means of getting from one place to the other in an energy efficient manner. While walking may seem like a simple task, it requires the coordination of multiple systems in the body. The musculoskeletal, cardiopulmonary, and nervous systems are all active during walking (Cha et al., 2016). Controlling locomotion also involves the organization of both feedforward motor patterns and a combination of neural and mechanical feedback (Dickenson et al., 2000). There are multiple levels regarding the neural control of movement, ranging from incoming sensory inputs to motor output, and it is necessary to study each of them to gain a more holistic understanding of locomotion.

While most studies mainly focus on studying the biomechanics of forwards walking, there is surprisingly not a lot of information about the biomechanics of backwards walking (van Derusen et al., 1998). There is some evidence that suggests that aspects of BW are just the time-reversed version of FW (Jansen et al., 2012). However, other aspects such as EMG activity of various antagonistic muscles are not the same when looking at time-reversed data (van Deursen et al., 1998). Given there are multiple levels that could have been affected, it is necessary to figure out which level of neural control caused this change. Studying these differences could give researchers more insight into the neural networks that control and activate different motor actions.

Neural Control of Walking

Walking is a complex task that requires the body to coordinate multiple different muscles to generate movement while maintaining stability mid-movement (Lacquaniti et al., 1999). Movement can be broken down into multiple levels, and one important level of motion is the innervation of information at the spinal cord. The spinal cord is broken down into multiple segments, each with pairs of nerves that activate specific skin areas or muscles (Guertin, 2013). Some of these segments included the thoracic segments, which help control motor control of the finger, and sacral muscles which are involved in coordinating locomotion (Guertin, 2013). Additionally, there are two different white-matter tracts in the spinal cord that help coordinate specific types of movements, the pyramidal and extrapyramidal tracts (Guertin, 2013). The pyramidal tract contains axons that are involved in coordinating skilled movements, while the extrapyramidal tract helps maintain postural control and general locomotion (Guertin, 2013). However, this is not the beginning of the descending pathway for locomotion. These tracts connect to different parts of the brain. The extrapyramidal tract originates in the subcortical nuclei, specifically the pons (Guertin, 2013). The pyramidal tract originates in the cerebral cortex as well as some brainstem motor nuclei, and it is one of the most direct descending motor pathways between the brain and the final motor component (Guertin, 2013). These white matter tracts serve as information relays between the brain and the spinal cord.

The gray matter tracts, on the other hand, are thought to be involved in reflex pathways as well as more complex circuits involved in stereotyped movements (Guertin, 2013). One type of neural network is thought to be a large driver for movement in many organisms, central pattern generators (CPG). A CPG is a neuronal circuit that can produce rhythmic motor patterns like walking without requiring a descending input (Marder, 2001). While CPGs can induce rhythmic activity without sensory information, a stimulus or modulator is often required to activate it (Marder, 2001). CPGs often consists of multiple pre-motor interneurons that help activate motor neurons directly connected to the CPG, but these motor neurons can also be part of the CPG itself (Marder, 2001). In rodent models, Guertin (2009) identified several new CPG neuron candidates located within the upper lumbar segments of the thoracolumbosacral spinal cord (Figure 1). V1 interneurons have been shown to be involved in high locomotor frequencies, and V3 interneurons were found to help create robust and balanced rhythm during movement (Guertin, 2009). It is thought that walking, specifically the flexion and extension of leg muscles, could be explained by rhythmic circuits like the CPG (Marder, 2001). The antagonistic leg muscles are activated by different motor neurons, meaning that any coordination between these muscles necessitates a connection between the two groups of neurons responsible for activating the opposing muscles (Lacquaniti et al., 1998)

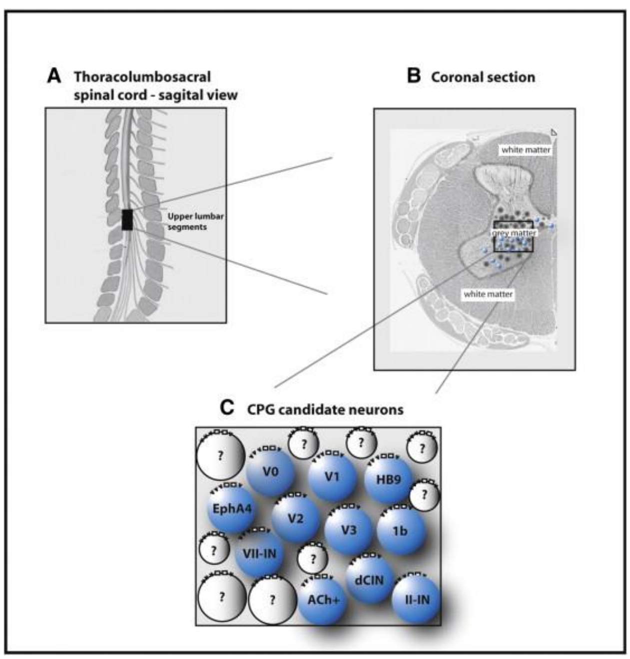

Figure 1: Potential CPG candidate neurons located within the spinal cord and their projections to the brain (Guertin, 2009)

While CPGs provide the underlying framework of a rhythmic pattern of activation, it does not explain how gait characteristics and movement change under different conditions. Motor movements are often adjusted to account for both neural and mechanical feedback, both of which are important in structuring CPG motor activity (Dickinson et al., 2000). The strength of the sensory feedback questions whether these sensory neurons should be seen as modulators of CPGs or as a part of the network itself (Dickenson et al., 2000). During locomotion, neural feedback from sensors comprise of three different inputs. The first is tonic inputs from directional sensors including the eyes and ears, and this affects speed, the direction of motion, and helps guide people to a specific direction while avoiding obstacles (Dickenson et al., 2000). The second form is from specialized equilibrium organs such as the inner ears, and this helps maintain postural stability as people move (Dickenson et al., 2000). The last input is rapid phasic feedback which comes from mechanosensory cells, and this is used to adjust cyclic motor patterns by stimulating cells within CPGs or by activating motor circuits that run in parallel with the specific CPG (Dickenson et al., 2000). Overall, these inputs help describe internal and external environmental changes, allowing the body to fine-tune motor output for each unique scenario (Dickenson et al., 2000).

Another way to look at human movement is by analyzing the kinematics of gait and the dynamics of the leg. There are several kinematic features, like the rotation of the pelvis, that aim to minimize energy expenditure during walking gait (Kuo et al., 2010). However, other features like reducing displacement during walking do not decrease energy expenditure (Kuo et al., 2010). By analyzing kinematics, researchers have come up with models that could explain our walking gait. One model is the inverted pendulum model which states that the stance leg acts as an inverted pendulum (Kuo et al., 2010). The “pendulum” leg will conserve mechanical energy, thus not requiring any additional mechanical work to keep producing gait (Kuo et al., 2010). If there is a change in kinetic energy, gravitational potential energy will offset it to prevent extra mechanical work being performed by the muscle (Kuo et al., 2010).

Models like the inverted pendulum model fit under the idea of dynamic walking which states that locomotion is produced by the passive dynamics of the leg even in the absence of active control (Kuo et al., 2010). Passive dynamics states that natural movements, including walking, occur due to the interaction of gravity, inertia, and elasticity (Gan et al., 2018). Gan et al. (2018) believes that natural mechanical dynamics of a bipedal system will induce passive dynamic gaits. Their results support the notion that gaits are generated by the action of natural mechanical dynamics of a legged system without the presence of control. In most passive dynamic models, cognitive controls, like sensory information, are excluded, allowing researchers to focus purely on the dynamic aspects of gait (Handžić et al., 2013).

Differences Between Forwards and Backwards Walking Biomechanics

The biomechanics of walking and running are characterized by two phases: stance and swing phase (Kharb et al., 2011). During stance phase, the foot is planted on the ground and acts as a pivot point. Swing phase starts when the foot is lifted off the ground (Dadashzadeh et al, 2017). Stance phase takes up approximately sixty percent of the gait cycle while swing phase takes up the last forty percent (Umberger, 2010). Stance phase can be broken down further based on whether there is support from one leg or both legs (Figure 2). There are intermediary periods between each phase known as take-off and touch-down. Take-off is the period between stance and swing phase when the foot is about to leave the ground, and touch-down is the exact time when the foot hits the ground again (Dadashzadeh et al, 2017). During this cycle, the leg is thought to move like a pendulum under the influence of gravity. However, while this does describe how the leg moves, it does not explain the more minute details regarding leg muscle activation during swing or stance phase, suggesting that each phase is controlled by some neuromuscular system (Umberger, 2010).

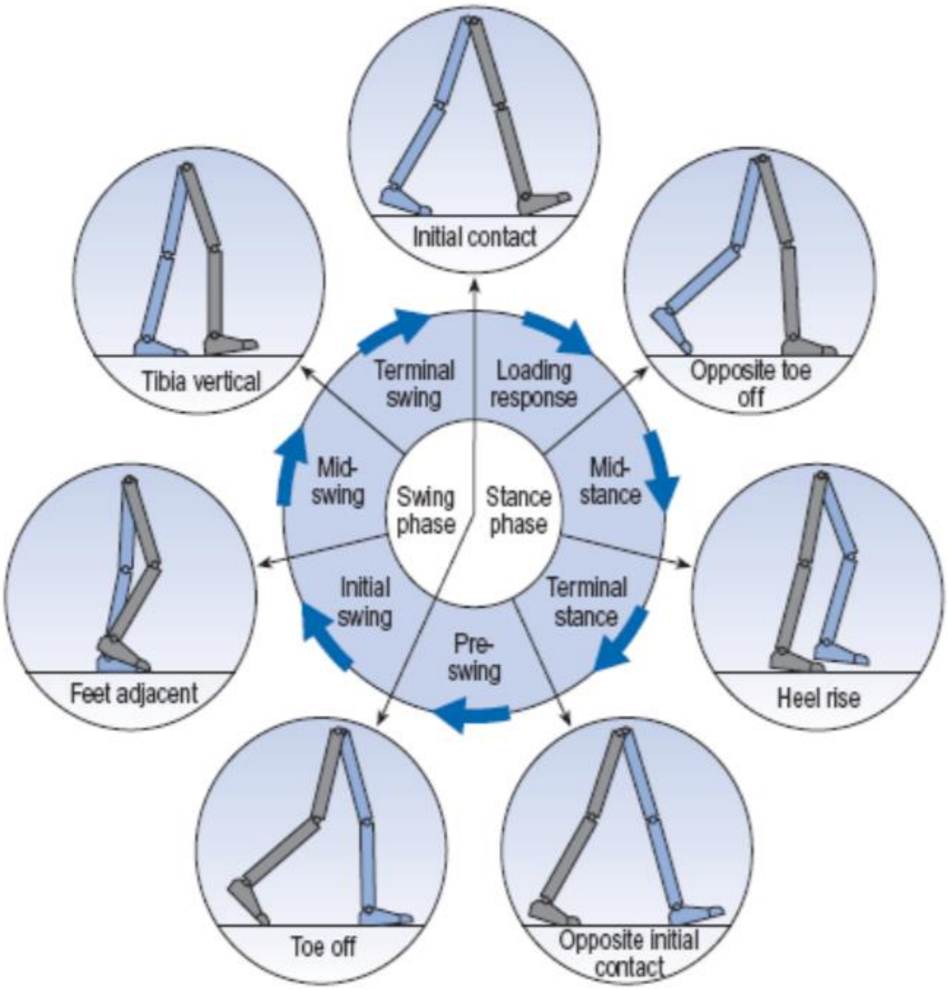

Figure 2: A breakdown of the different events that define both stance and swing phase (Kharb et al., 2011).

Motion is usually generated by the anterior and posterior muscle groups of the lower legs (van Deursen et al., 1998). These muscles are reciprocally activated based on the two phases of gait (van Deursen et al., 1998). Two antagonistic muscles that are activated during stride are the tibialis anterior (TA) and the gastrocnemius lateralis (GM). Plantarflexion of the GM is seen during stance phase (van Deursen et al, 1998). In EMG recordings, a burst of activity in the GM is usually seen during the early phases of stance (foot plant), then returns to baseline midstance, and ends with a small burst of activity during late stance when the toe lifts off the ground (Grasso et al., 1998). On the other hand, we see dorsiflexion of the TA during swing phase, and it is activated when the leg folds in the air (van Deursen et al., 1998). In EMG recordings, it was found that the TA is activated during the beginning and end of swing phase, and its activity returns to baseline mid-swing (Figure 3).

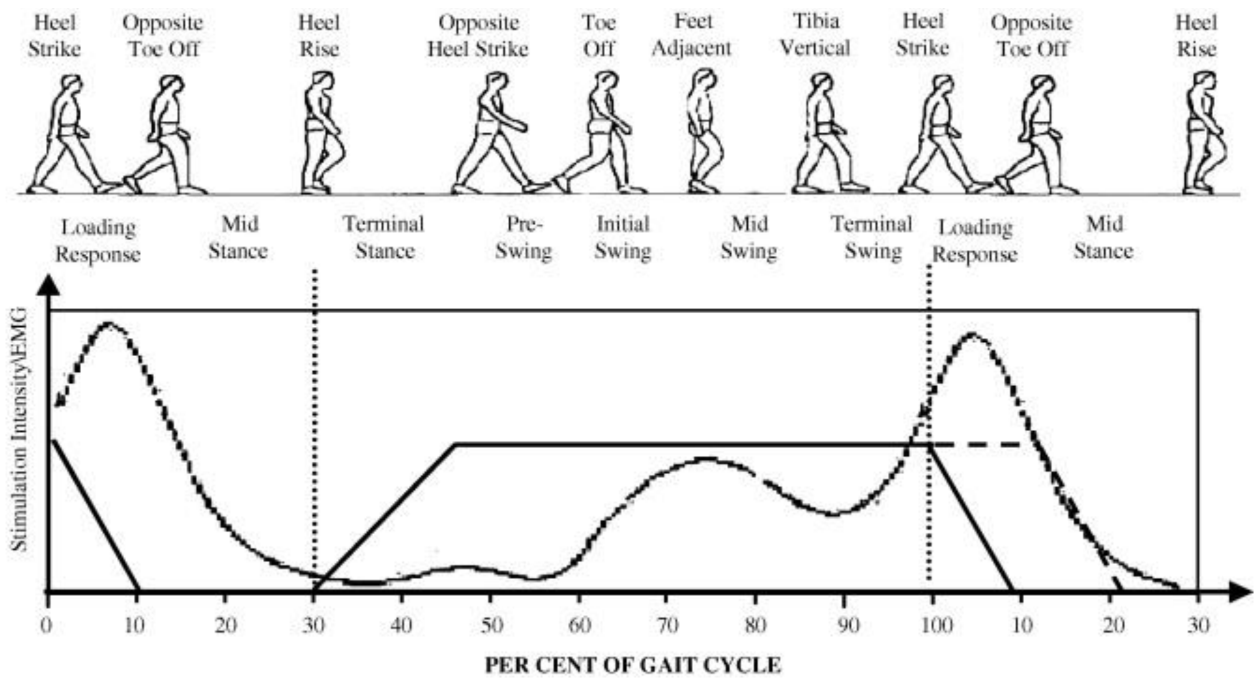

Figure 3: EMG pattern for TA shows high levels of activation during both initial swing and terminal swing (Byrne et al., 2007).

While these models describe forwards walking, it is important to consider how they can also be applied to backwards walking (BW). There are not many human motions that can be reversed, but studying reversible walking may provide a larger insight into motor pattern representations for locomotion (Grasso et al, 1998). When researching BW, it is mainly studied using CPGs. It is thought that BW gait could be recreated by switching the coupling of unit CPGs for a specific limb to perform regular motion in reverse (Grasso et al., 1998). Some aspects of BW can be compared to forwards walking (FW) such as hip and ankle joint motions after time-reversing the data.

In BW, similar muscles are used to generate movement, but what defines the two phases of gait differ. BW stance phase begins when the toes touch the ground as opposed to the heel touching the ground in FW, and swing starts when the heel lifts off the ground (Grasso et al., 1998). This suggests that muscle activation between FW and BW will be slightly different as well. For example, the GM in FW was used to start movement, while the GM in BW changes function and helps slow down movement when the toes hit the ground (Cipriani et al, 1995). In FW, GM activity was recorded during the early and late stance, while the GM activity was recorded during midstance in BW (Grasso et al, 1998). While the timing of muscle activation differs between BW and FW, the time spent in stance and swing phase is very similar (van Deursen et al., 1998).

However, it is still unclear whether the muscle activation for FW and BW is the same (Lee et al., 2013). Some surface level EMG research suggests there is similar muscle activation in FW and BW when you time-reverse the BW data for upper-leg muscles (Jansen et al., 2012). From a neural control perspective, it would be advantageous for a single muscle to have multiple functions such that it could be used in both FW and BW (Jansen et al, 2012). While this would be convenient, EMG patterns do not support this hypothesis. EMG patterns in BW are considerably different than the profiles seen in FW, suggesting that changes need to be made to normal locomotor function to account for this change in direction (van Deursen et al., 1998). van Derusen et al. (2018) found that there are phase shifts (around 25%) in muscle activation patterns in four out of six antagonistic leg muscles, suggesting that an adaptation in the mechanism of FW occurs, not necessarily reversing the mechanism. Jansen et al. (2012) also suggests that

CPGs could be altered to create different locomotive behaviors. One CPG does not necessarily use the entire network for FW, but instead, it uses part of it to help create the motion required for BW (Jansen et al., 2012).

Current Knowledge regarding FW and BW Gait Differences.

Most studies conducted had patients walk in an open environment where they could control their speed. In other words, they did not walk at the same speed in FW and BW. There are noticeable decreases in speed and stride length in BW compared to FW (Lee et al., 2013). This decrease in speed can be attributed to a higher difficulty regarding BW. The lack of visual cues and higher degree of instability make BW more demanding than FW. Not only that, BW is not a stereotyped movement and is less comfortable than FW, meaning that the participants could have been afraid of falling, thus leading to a decrease in speed (Cadenas-Sanchez et al., 2015). In terms of stance and swing phase, Lee et al. (2013) found that there was no significant difference in stance or swing phase percentages.

With regards to neural control and EMG data, recent research suggests that muscle contributions in FW and BW can be time-reversed to show a similar pattern of horizontal acceleration (Jansen et al., 2012). The same muscles were found to be involved in horizontal movement during FW and BW, and this is important in showing that these muscles can be used to achieve movement in the opposite direction. There was a theory that thought that neural control for reversed movement was separated into different systems, but these results help simplify neural control by demonstrating that the same mechanisms are used for both regular and reversed movement (Jansen et al., 2012)

How Cognitive Tasks Could Interfere with Walking Gait

Given the modern era of technology, multitasking is become more common. Studies have shown that using a laptop or cell phone while sitting in class greatly limit recall of class material (Schott et al., 2018). The ability for an individual to do multiple tasks at once while performing equally well on both requires a large amount of energy. This problem brings up questions regarding dual-tasking specifically related to motor and cognitive tasks (Schott et al., 2018). Researchers are interested in how performing two tasks at the same time affect performance on each task. Both tasks require cortical resources and interfere with attention, and this phenomenon is known as cognitive-motor interference, or CMI (Schott et al., 2018).

CMI causes people to adopt different strategies to perform both tasks at once, potentially leading to worse performance in both tasks (Bradford et al., 2019). The brain must make a strategy to allocate resources efficiently in order to perform both tasks at the same time (Bradford et al., 2019). One theory suggests that all tasks, not limited to cognitive or motor tasks compete for cortical resources, leading to a reallocation of said resources to perform one task at the expense of the other (Bradford et al., 2019). For example, the motor task could use more resources in order to maintain balance and walking gait regularity while doing poorly on the cognitive task. Alternatively, the cognitive task could take priority and lead to gait asymmetry. It is not just the type of tasks involved that affect walking gait, but also attentional resources, age, and neurological comorbidity (Janssen et al., 2019).

Walking gait is a rhythmic action with little variation, but the introduction of environmental distractions and additional tasks could cause gait to change considerably (Schaefer et al., 2015). These distractions may also lead to an imbalance in postural stability, or the ability to keep the center of mass steady while the base of support is constantly changing

(Priest et al., 2008). Studies suggest that attention span plays a large role in the regulation of gait, but this depends greatly on the age of the individual. It is thought that there are larger amounts of CMI in younger children and older adults, leading to more erratic walking gait (Schaefer et al., 2015). Higher levels of CMI are also found in adults that have clinical conditions such as Parkinson's disease, dementia, or a concussion (Schott et al., 2018).

CMI's effects on walking gait depend heavily on the individual. When given a cognitive task, most people often show a decrease in speed and/or worsening in cognitive task performance (Timmermans et al., 2018). When looking at stroke-patients or older adults, CMI is often associated with increases in falling risk. However, not all individuals experience the same gait interferences. These effects are also prominent in patients with advanced stages of Parkinson's disease (Janssen et al. 2019). A response to being given a cognitive task is to walk slower in order to shift resources to perform that new task while still maintaining balance to walk (Timmermans et al., 2018). When studying this, the decrease in speed is generally controlled by placing subjects on a treadmill. It might be more imperative to analyze how people change task preferences when walking to gain a better understanding of how dual-tasking and task prioritization work (Timmermans et al., 2018). In other words, it would be interesting to see how individuals, when allowed to slow down, change task preferences and how the new information interferes with descending motor inputs from the brain.

Types of Cognitive Tasks and CMI's effect on Gait

There are a large variety of cognitive tasks that researchers can use when testing CMI. There are four common tests that were used in these studies (Al-Yahya et al., 2011). These involve reaction time tasks, discrimination and decision-making tasks, mental tracking, and verbal fluency tasks. Each of these tests is meant to analyze different aspects of cognition ranging from processing speed to response inhibition in terms of the cognitive task and locomotion (Al-Yahya et al., 2011). There is evidence from brain imaging studies that suggests that areas related to cognitive functioning are activated during actual, imagined, or simulated gait (Al-Yahya et al., 2011). This suggests that, while CPGs are at play and control most muscle activation, there can be still be external factors that interfere with rhythmic like locomotion. Overall, many studies have found that CMI tends to lead to decreased speed, cadence, stride length, and increased stride time and stride variability (Al-Yahya et al., 2011).

The brain, as stated previously, has a limited number of resources to process multiple tasks (Bradford et al., 2019). The brain constantly takes in different sensory inputs and integrates that information in order to allow the body to move in a safe and efficient manner. Dickenson et al. (2000) notes how there are different inputs involved in fine-tuning specific aspects of movement, including speed and postural stability. When the cognitive task is presented, it could compete with sensory stimuli for cortical resources, explaining why dual tasking makes walking more difficult and irregular. The cognitive task would interfere with sensory processing and disrupt the downstream motor pathway, leading to irregular gait and higher risks of falling.

Experiment

A large amount of research today focuses primarily on cognitive motor interference during forwards walking (for a review see Al-Yahra et al., 2011). However, there is a lack of research on how CMIs affect BW. There are differences in muscle activation and stride characteristics between FW and BW, but how would these differences change if a cognitive load were introduced? Would there be a greater difference between the gait characteristics, or would there just be larger variance among the data overall?

The goal of this study was to determine the effect of a cognitive load on forward and backward walking and see if downstream motor processing would be affected by the cognitive load. Subjects walked on a treadmill while performing a cognitive task, and they were given the cognitive task during both FW and BW. We hypothesized that gait characteristics are more variable during backward walking, with greater variability among the conditions that experienced a cognitive load and BW with a cognitive load will vary the most. Given the cognitive task will interfere with sensory inputs for cortical resources, it will prevent the sensory information from being processed and lead to more postural instability while walking backwards. Because of this, subjects may adopt different inefficient strategies to maintain balance that would normally be corrected if sensory feedback were processed correctly. There is also the complete lack of visual feedback regarding where the subject is walking. The subjects will have to rely more on other forms of sensory feedback to compensate, thus leading to more variability in gait.

Methods

Participants

Fifteen young adults (male and female) volunteered to participate in the experiment. The age of participants ranged from 18 - 22 years of age. All participants were comfortable with walking forwards and backwards on a treadmill. Two subjects' data was noisy due to the inability to consistently track the position data in one or more experimental conditions, so data analysis for them was not possible. Because of this, data analysis was performed on thirteen participants. All procedures were approved by the CMC IRB committee.

Experimental Set-Up

Seven reflective markers were placed on each participant's right leg: 2 on the inside and outside of the knee, ankle, and heel each, and 1 on the toe. These markers were tracked by three Oqus cameras placed at different locations around a treadmill on which the subjects walked and recorded in the Qualisys Track Manager (QTM) software. Each trajectory recorded the time and position data for each individual marker. A complete stick figure connecting markers to create each joint was created in QTM software.

Procedure

Before beginning data collection, participants were familiarized with a cognitive task. The cognitive task was an arithmetic subtraction task of which they had to count backwards aloud from a randomly generated three-digit number in intervals of a number between 6 to 9 (e.g. counting backwards from 150 in intervals of 7). The cognitive task was the same for all trials, but a different three-digit number and interval was given for each trial and participant. Participants were then asked to walk on a treadmill at a speed of between 1 to 3 mi/hr. To identify their preferred speed, participants walked at different speeds and chose one at which they could walk forwards and backwards comfortably without stumbling. There were four walking conditions which were performed in a randomized order: forwards walking (FW), backwards walking (BW), FW with the cognitive task, and BW with the cognitive task. Each walking condition lasted 30 seconds (approximately 10-20 steps), and ample time was given in between each condition for familiarization. Gait data was recorded once participants reached steady-state walking.

Data Analysis

The dependent variables for QTM were the following: stance phase percentage, swing phase percentage, and stride length (mm). For each stride, the stride time will be measured by looking at the time difference of the toe and heel marker. The stride time was used to calculate the stance and swing phase percentages. The start time was defined as when the heel initially touches the ground, and the end time will be when the heel touches the ground again.

This method was also used to measure the stride length. We used the position data for the heel at the start and end times, and we took the difference between them to get the stride length. With regards to stance and swing phase percentages, the stance phase for FW will be characterized by time-stamp of initial heel contact on the ground to toe lift-off from the ground. The swing phase for FW is characterized by the period of time when the right leg is off the ground. The stance and swing phases for BW is just the time-reversed pattern of FW. Data analysis was done on six steps per experimental condition. To analyze data across all subjects, stride length needed to be standardized due to differences in leg length between the participants. To account for this, a ratio of leg length (mm) to stride length (mm) was made. Stance and swing phase were already standardized based on stride time for each subject.

When analyzing data within each subject, a two-way, repeated measures ANOVA was used to determine both the interactive and individual effects of direction and cognitive load on specific walking gait characteristics. R software was used to analyze the effects across the entire subject pool. If significant, a Tukey’s HSD test was run to examine differences between experimental conditions across the entire subject pool.

Results

Subject 4 was chosen to represent individual variation since the variance between trials was low (Table 2). While this subject's data, is not representative of the whole subject pool (Table 1), it demonstrates how a single individual changed across conditions.

Population Statistics

Stride Length

Across the entire subject pool, stride length was significantly shorter during all backward walking compared to all forwards walking ($p < 0.001$). However, stride length was not different between trials with and without a cognitive load, and there was no significant interaction (Figure 4).

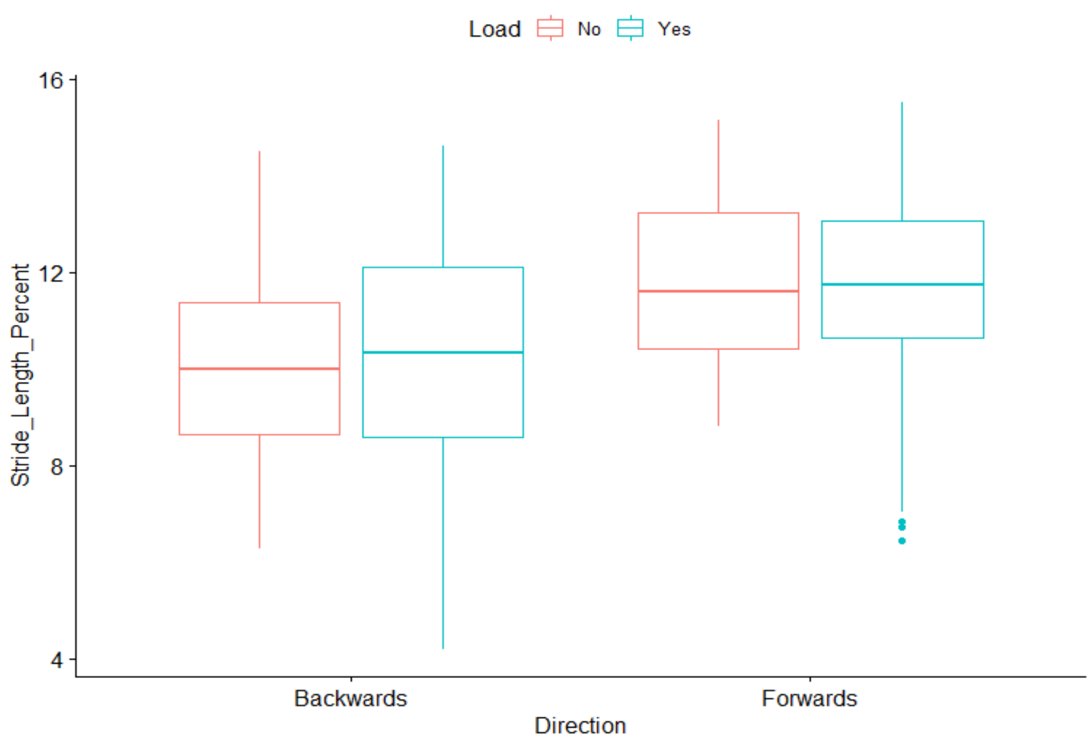

Figure 4: Stride Length differences (leg length/stride length) between 4 experimental conditions across all subjects (n=13). Stride length was significantly shorter during backward compared to forward walking.

Stance Phase Percentage

The data was not normalized for the entire subject pool, so a square root transform was used (Figure 5). Stance phase in all backwards walking trials was significantly shorter than stance phase in forwards walking trials (p = 0.029). However, stance was not significantly different between trials with and without a cognitive load, and there was no interaction effect (Figure 6).

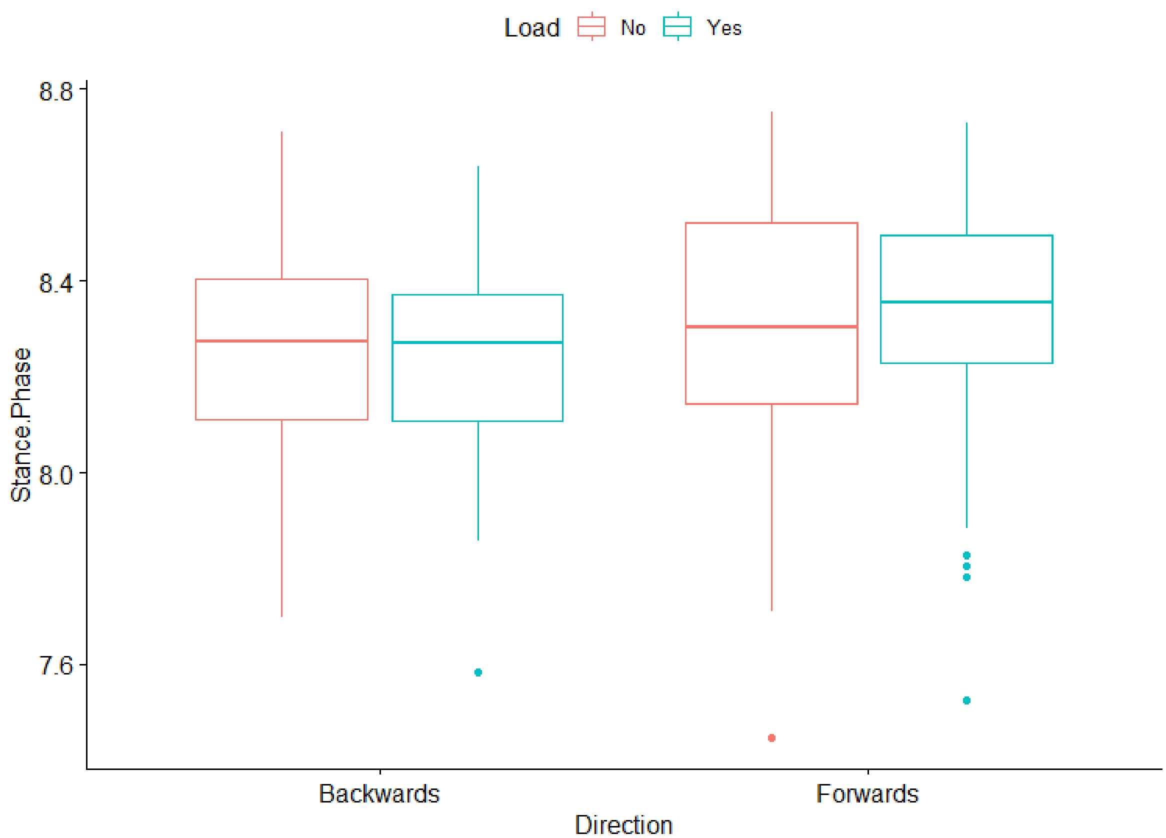

Figure 5: Stance Phase (square root transformation) differences between 4 experimental conditions across all subjects (n=13).

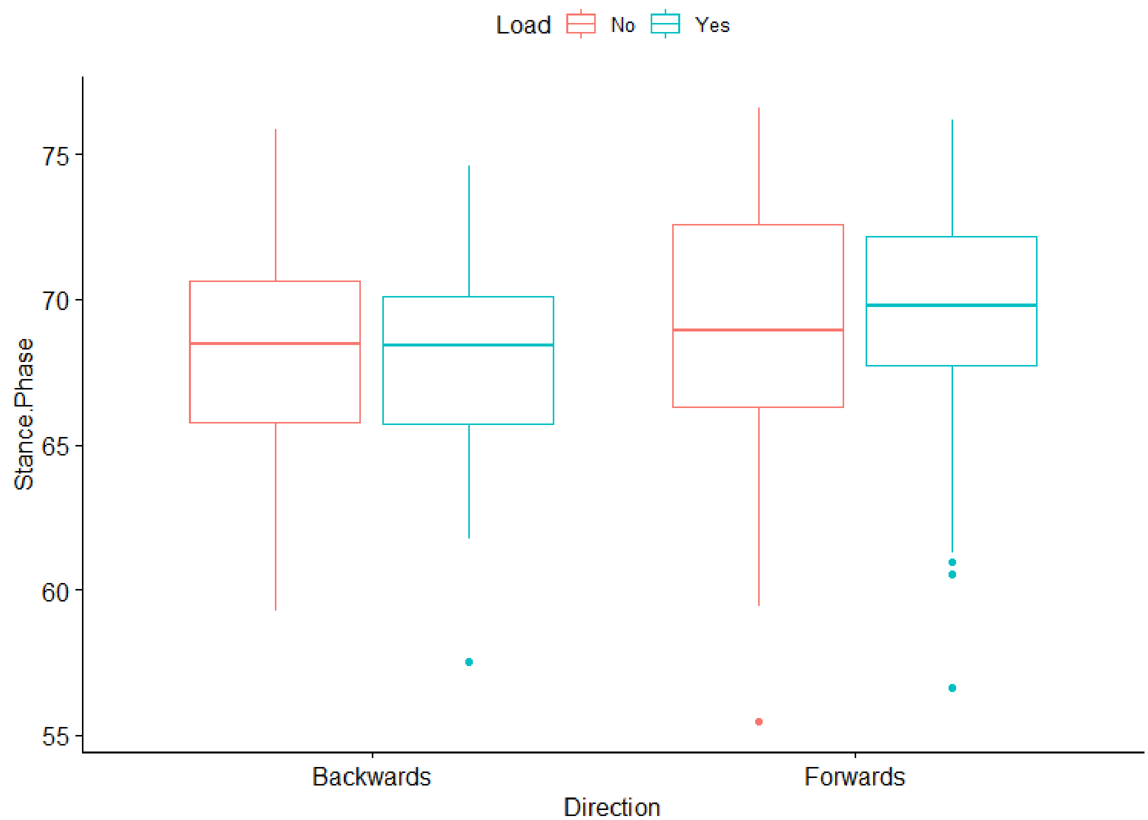

Figure 6: Stance Phase (raw) differences between 4 experimental conditions across all subjects (n = 13) Stance phase was longer in forwards compared to backwards walking.

Swing Phase Percentage

Before analyzing data across all subjects, a log transform was used to normalize the data (Figure 7). Swing phase in all backwards walking was significantly larger than all forwards walking ($p = 0.015$). However, swing phase was not different between trials with and without a cognitive load, and there was no significant interaction (Figure 8).

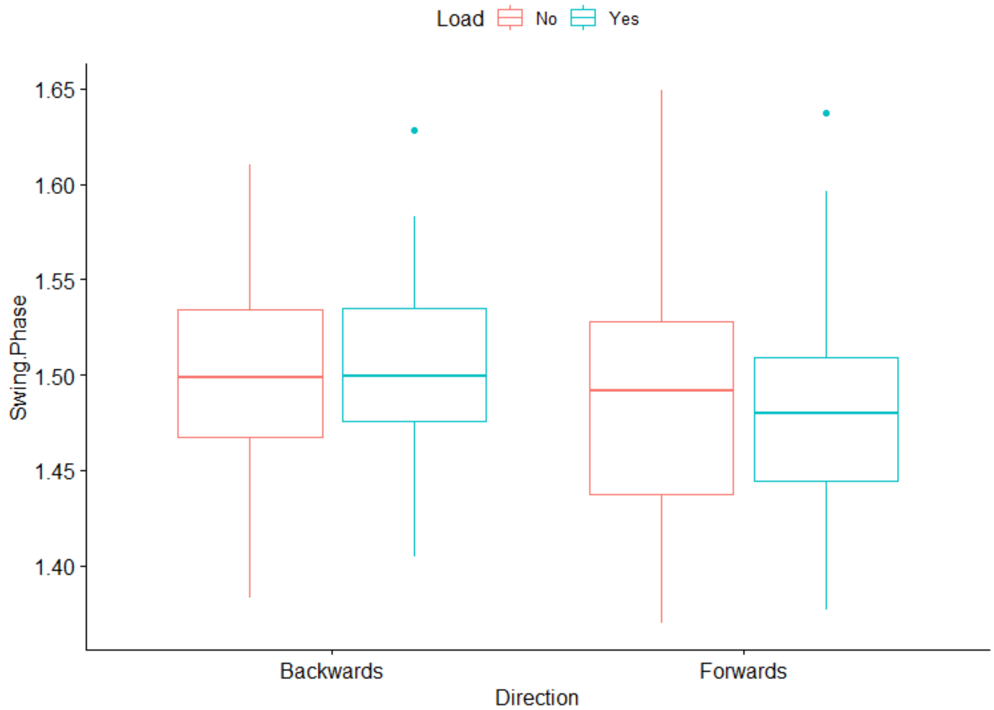

Figure 7: Swing phase (Log transformation) differences between 4 experimental conditions across all subjects (n = 13)

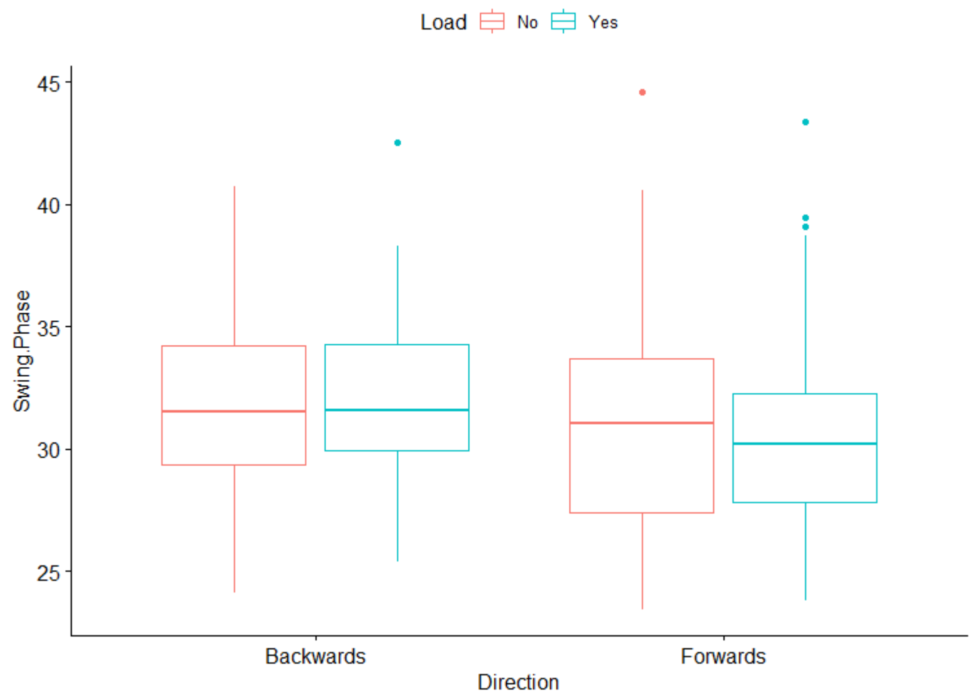

Figure 8: Swing Phase (raw) differences between four experimental conditions across all subjects (n = 13). Swing phase is significantly larger in backwards compared to forwards walking.

Individual Statistics

Stride Length

All of subject 4's backwards stride length was significantly smaller than all forwards stride length (p = 0.001). Stride length in all no-load conditions is much smaller compared to with load conditions (p = 0.004), but no interaction effect was observed (Figure 9). When comparing stride length between the subjects, 9 of the 13 subjects exhibited greater stride lengths during forwards walking, 1 of the 13 subjects exhibited greater stride lengths during backwards walking, and the last 3 subjects showed no change in stance. With regards to cognitive load, 3 of the 13 subjects exhibited greater stride lengths when the cognitive load was present, 1 subject exhibited greater stride lengths when it was not present, and 9 of the 13 subjects showed no significant difference (Table 2).

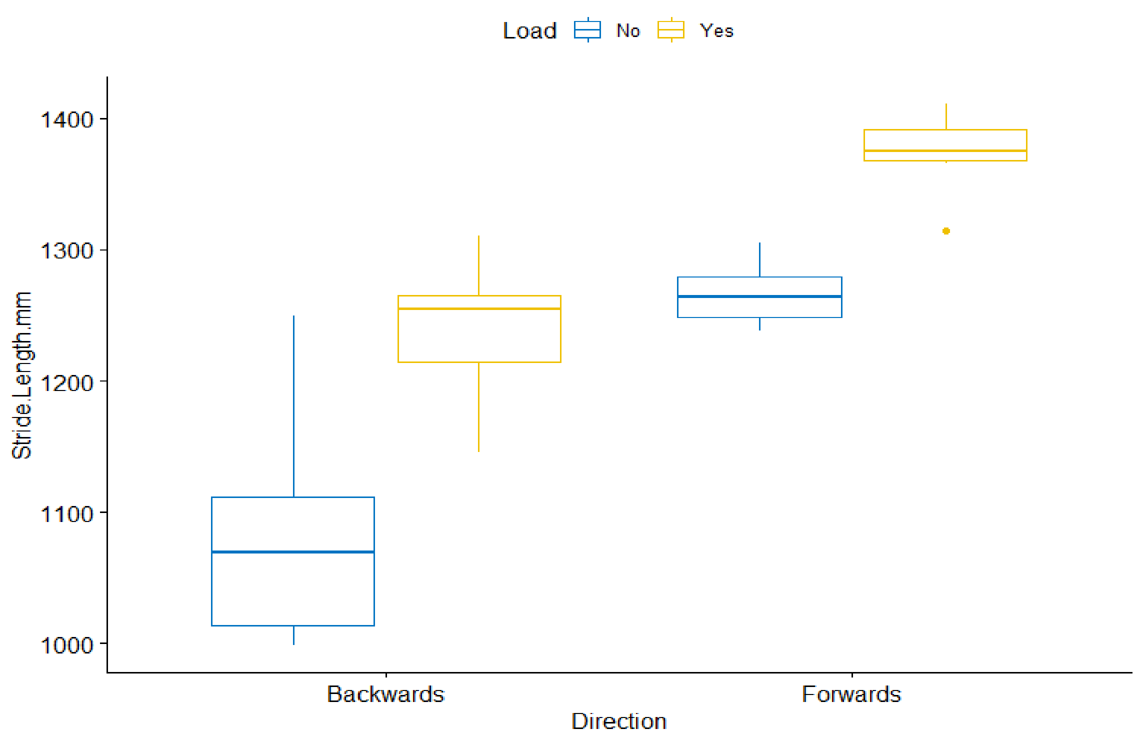

Figure 9: Stride Length (mm) differences between four experimental conditions for one subject (n = 24)

Stance Phase

Stance duration for all backwards walking trials was significantly smaller than all forwards walking trials ($p < 0.001$). As for cognitive load, all trials without the cognitive load showed much shorter stance duration than trials with the cognitive load ($p < 0.001$), and there was a significant interaction effect ($p < 0.001$) (Figure 10). When comparing direction, 5 of the 13 subjects exhibited larger stance duration in all forwards walking trials, 2 of the 13 subjects showed larger stance duration in all backwards walking trials, and the remaining 6 showed no significant difference. For cognitive load trials, 2 of the 13 subjects exhibited longer stance duration in all trials with the cognitive load, none of the subjects shoed longer stance duration in all trails without the cognitive load, and the remaining 11 subjects showed no significant difference (Table 2).

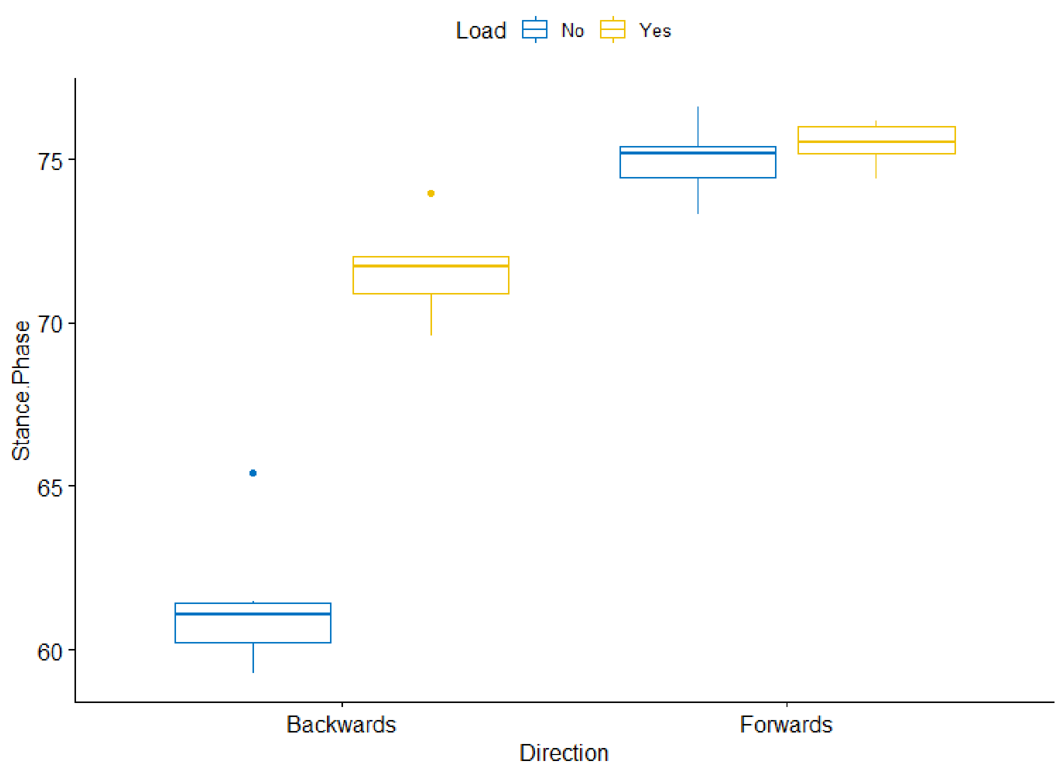

Figure 10: Stance Phase differences between four experimental conditions for one subject (n = 24)

Swing Phase

Swing phase in all backwards walking trials was significantly larger compared to all forwards walking trials ($p < 0.001$). Swing phase was also much larger in all trials without the cognitive load than when the cognitive load was presented ($p < 0.001$), and a significant interaction effect was observed ($p < 0.001$) (Figure 11). When comparing direction, 5 of the 13 subjects exhibited longer swing duration in all backwards walking trials, 2 of the 13 exhibited longer swing duration in all forwards walking trials, and the remaining 6 showed no significant difference. As for the presence of cognitive load, 2 of the 13 subjects showed longer swing duration during all trials without cognitive load, none showed longer swing duration during trials with the cognitive load, and the remaining subjects showed no significant difference (Table 2).

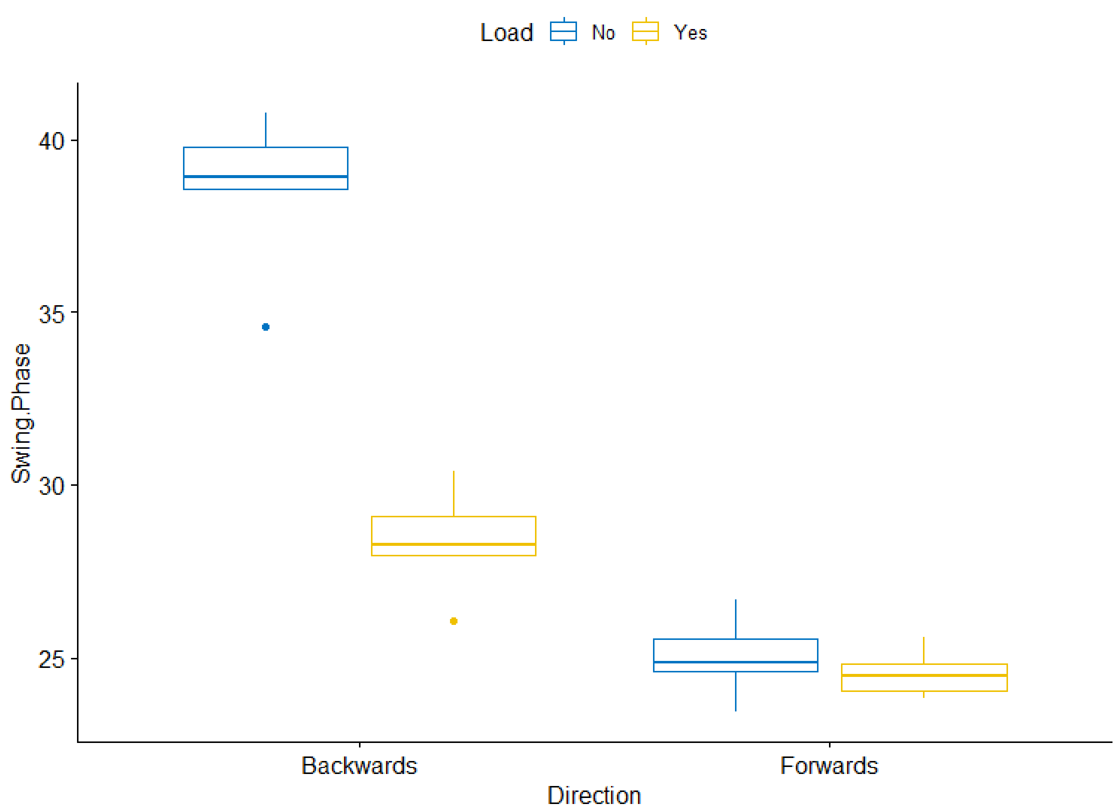

Figure 11: Swing Phase differences between four experimental conditions for one subject (n=24)

Table 1: Differences in Gait Characteristics across the entire subject pool (n = 13)

Gait Characteristic	Direction		Cognitive Load		Interaction	
	p-value	t-value	p-value	t-value	p-value	t-value
Stride Length	**< 0.001**	23.089	0.100	1.649	0.179	- 1.347
Stance Phase	**0.029**	2.200	0.879	0.152	0.654	0.449
Swing Phase	**0.015**	-2.442	0.996	0.005	0.628	- 0.485

Table 1: Differences in Gait Characteristics for each subject

Subject	Gait Characteristics	Direction		Cognitive Load		Interaction	
		p-value	F-value	p-value	F-value	p-value	F-value
1							
	Stride Length	**0.049**	6.708	0.643	0.243	0.925	0.010
	Stance Phase	0.409	0.813	0.101	4.039	**< 0.001**	81.992
	Swing Phase	0.409	0.813	0.101	4.039	**< 0.001**	81.992
2							
	Stride Length	**< 0.001**	71.318	0.787	0.081	**0.014**	13.595
	Stance Phase	**0.003**	21.194	0.248	1.705	0.891	0.021
	Swing Phase	**0.003**	21.194	0.248	1.705	0.891	0.021
4							
	Stride Length	**0.001**	46.018	**0.004**	25.443	0.452	0.664
	Stance Phase	**< 0.001**	318.011	**< 0.001**	55.908	**< 0.001**	60.822
	Swing Phase	**< 0.001**	318.011	**< 0.001**	55.908	**< 0.001**	60.822
5							
	Stride Length	**< 0.001**	814.942	0.835	0.048	0.117	3.592
	Stance Phase	0.263	1.586	0.164	2.657	0.314	1.250
	Swing Phase	0.263	1.586	0.164	2.657	0.314	1.250
6							
	Stride Length	0.69	0.178	**0.009**	17.193	0.619	0.280
	Stance Phase	**< 0.001**	56.483	**0.03**	9.017	**0.002**	33.222
	Swing Phase	**< 0.001**	56.483	**0.03**	9.017	**0.002**	33.222
7							
	Stride Length	**0.012**	15.095	**0.012**	14.893	0.082	4.697
	Stance Phase	0.283	1.449	0.063	5.647	0.964	0.002
	Swing Phase	0.283	1.449	0.063	5.647	0.964	0.002
8							
	Stride Length	**0.006**	20.848	0.128	3.318	0.166	2.629
	Stance Phase	**0.024**	10.202	0.191	2.285	0.803	0.069
	Swing Phase	**0.024**	10.202	0.191	2.285	0.803	0.069
10							
	Stride Length	0.974	0.001	0.672	0.201	0.805	0.068
	Stance Phase	0.193	2.259	0.979	0.0007	0.213	2.037
	Swing Phase	0.193	2.259	0.979	0.0007	0.213	2.037
11							

	Stride Length	0.303	1.318	0.453	0.660	0.667	0.208
	Stance Phase	0.555	0.399	0.065	5.565	**0.014**	13.760
	Swing Phase	0.555	0.399	0.065	5.565	**0.014**	13.760
12							
	Stride Length	**0.002**	38.722	< **0.001**	82.810	**0.023**	10.550
	Stance Phase	0.076	4.974	0.068	2.634	0.421	0.767
	Swing Phase	0.076	4.974	0.166	2.634	0.421	0.767
13							
	Stride Length	< **0.001**	190.699	0.119	3.521	**0.009**	17.067
	Stance Phase	**0.002**	34.081	0.106	3.883	0.515	0.489
	Swing Phase	**0.002**	34.081	0.106	3.883	0.515	0.489
14							
	Stride Length	**0.044**	7.117	0.386	0.902	0.808	0.066
	Stance Phase	**0.010**	16.270	0.450	0.670	0.679	0.192
	Swing Phase	**0.010**	16.270	0.450	0.670	0.679	0.192
15							
	Stride Length	< **0.001**	117.844	0.904	0.016	0.273	1.515
	Stance Phase	**0.004**	25.915	0.790	0.079	0.089	4.427
	Swing Phase	**0.004**	25.915	0.790	0.079	0.089	4.427

Discussion

The results of our experiment are quite similar to previous studies examining the differences between FW and BW gait characteristics. Direction had a significant effect on all gait characteristics analyzed. This was seen by the noticeably smaller stride length in all BW trials compared to FW trials. The average for all FW stride length ratios was greater than BW by approximately 2% (Figure 1). Stance duration was also much longer in FW trials than BW trials by approximately 1.5% (Figure 2). Swing duration was also affected and was longer in BW trials than FW trials by the same percent as in stance duration (Figure 3). The results are like Lee et al.'s (2013) with regards to stride length, but they differ on stance phase. Lee et al. (2013) found that there was no significant difference in stance phase unlike our experiment.

There is a lack of research examining the effects of dual-tasking and CMI on other gait characteristics, including stance and swing duration, in both FW and BW. Many past studies, including Timmermans et al. (2018) and Schaefer et al. (2015), only analyzed the effects of CMI on FW and not BW. Another key feature of these studies is that they compared walking speed in FW and BW, something that was kept constant in our experiment. However, both Timmermans et al. (2018) and Schaefer et al. (2015) found that the presence of a cognitive task significantly affected walking speed. While our results do not contradict these past studies, it would be beneficial to examine CMI's effects on multiple gait characteristics, not just walking speed, in both directions.

It was interesting how only direction influenced the gait characteristics that were analyzed. In BW, the lack of tonic input from the eyes decreases the amount of sensory feedback, making the individual adjust their speed (Dickenson et al., 2000). Because speed was kept constant, it was expected that subjects would adjust their gait in other ways, and this was seen in the experiment. This suggests that CPG activity was modulated to account for a decrease in sensory feedback.

We believed that the cognitive task would compete for cortical resources and make it harder for the brain to process all the sensory information, but our results did not support this hypothesis (Bradford et al., 2019). We predicted that the introduction of the cognitive task would interfere with downstream processing of sensory information that helps motor processing. However, the cognitive task may not have had an effect due to the intrinsic nature of CPGs and how they are able to produce rhythmic, motor actions without requiring sensory stimulation (Marder, 2001).

The results also demonstrated that there was no significant interaction between direction and the presence of a cognitive task on gait characteristics. These results may have been attributed to a couple factors. The first was the order in which the cognitive task was presented.

The task was first introduced to the subject while sitting down, and then they were given a different number when they began walking. However, the participant could continue walking on the treadmill before being given the task. If walking began before presentation of the cognitive task, the CPG will already be producing the desired motor output and be relatively unaffected by the cognitive task due to the inherent nature of CPGs. If this were the case, then direction should be the only factor to influence gait characteristics since the CPG will no longer require downstream information processing to produce efficient and safe gait. This would mean that the cognitive task, though it might interfere with sensory processing, will have no effect on gait. Passive dynamics could also explain this. If walking already began, gait could already be primarily controlled by inherent mechanical dynamics of the leg, thus limiting any interference from the cognitive task on these dynamics (Gan et al., 2010). It would be interesting to see if the participant were asked to start the cognitive task and walking at the same time or even when asked to perform the cognitive task first.

The second factor was the constant speed during FW and BW. In many experiments, FW and BW data was recorded in an open environment where they could adjust their speed accordingly. While our results found that there were differences in stance and swing phase percentage between FW and BW, there are mixed reviews stating that there is no significant difference between directions (Lee et al., 2013). It was reported that individuals, when walking backwards and performing the task, tend to slow down due to a fear of falling (Cadenas-Sanchez et al., 2015). It can be suggested that they do this in order to maintain a normal walking gait while still performing both tasks well. However, this can be said both with and without the cognitive task. While there are slower speeds that are attributed to BW, this decrease in speed can explain the similarities in stance and swing phase percentages between FW and BW (Lee et al., 2013). Our experiment, unlike the previous ones mentioned, was performed on a treadmill. Participants were walked at a constant speed backwards, preventing them from regulating their gait. This could explain the significant difference of direction on the gait characteristics that were analyzed.

The last factor is familiarity with the cognitive task itself. In other words, the subject could remember the task and make strategies to perform the task without expending a large amount of cortical resources. Each participant was given the task a total of three times. While a different number and interval were given each trial, the subject could have figure out a pattern after a specific number of answers. This would decrease the amount of time they spend on the cognitive task and allow the subject to focus more on processing sensory information to maintain gait. In the future, it would be helpful to examine CMI's effects on gait characteristics using a cognitive task that subjects cannot become familiar with over time.

With regards to variation between subjects, not all individuals responded to the change in direction and cognitive load the same way. The gait characteristics affected by each variable between each subject were not the same. The different responses to each condition, shown by changes in gait characteristics, reveal that there is no definite answer for how people adapt to different environments. While our results show most subjects have shorter strides during backwards walking, this is not a universal trend. There were some subjects that showed no change in stride between FW and BW, demonstrating that each subject utilizes a different strategy in order to account for the different conditions. This suggests that each subject integrates sensory feedback differently based on both their own personal experiences and other environmental inputs, creating a unique motor output. While the nature of CPGs may remain the same across individuals, the motor output it produces will change based on how each individual processes sensory and mechanical feedback. This further illustrates the importance of downstream information processing on fine-tuning CPG activity to ensure safe, energy-efficient movement (Dickenson et al., 2000).

Future Direction and Implications

The most interesting facet of this research is task prioritization. Depending on what task the individual deems more important, they will devote more attention and resources to doing that task well at the expense of the other. However, it is interesting to see if what environmental conditions will attract or detract attention. Some studies found that the addition of a physical load (heavy weights) caused subjects to focus more on walking over the cognitive task (Bradford et al., 2019). While adjusting the order in which the cognitive task is presented, it would be interesting to do an experiment that compared gait characteristics when both a cognitive and physical load are present. Cognitive load is thought to detract attention away from walking and the physical load is thought to draw attention to it (Bradford et al., 2019). It would be interesting to determine if there is an interaction between the cognitive and the physical load and whether one has a stronger effect on attention and walking overall. One field that stems from this is analyzing strategies for adjusting movement under different conditions and how neural control plays a role into each strategy. For example, it would be interesting to analyze the strategies that underlying a decrease in stride and variations in stance/swing phase and the neural control mechanisms that make these strategies happen.

While our study did not look at the accuracy of the subjects when performing the cognitive task, their accuracy should be recorded as well to see how well they are performing relative to their walking performance. We told the participants to keep responding as they walked, but their responses were not recorded. While gait characteristics were measured between load and no-load conditions, the accuracy of the task itself may give a better insight as to how much each participant focused on the cognitive task compared to walking. In other words, future studies could examine the effects of walking on the cognitive task as well as vice versa. This may give researchers a better representation of how task representation takes place and how the motor task and the cognitive task interact with the performance of the other.

Another future study that can be conducted is based on one of the limitations of our study: its area of focus. Our study mainly focused on the effects of CMI and dual-tasking on FW and BW gait characteristics related to kinematics, but it did not analyze any features of gait kinetics, including joint angles or ground-reaction forces. It is important to analyze these features as well, for they may explain why specific gait kinematic differences were present.

Continuing to research BW could give scientists a better understanding of neural control mechanisms for locomotion. BW can be used as a form of therapy, for it helps enhance balance and strengthens lower limb muscles (Lee et al., 2013). With regards to rehabilitation for lower extremity injuries, BW can provide more benefits than FW. Some believe that it is a better form of therapy because it forces the individual to use more muscle activity with respect to effort than FW (Cipriani et al., 1995). BW also helps hemiplegic patients regain normal motion control and gait patterns. By studying BW, a derivation of FW, it could greatly increase understanding of how humans are able to control their own locomotive behavior (Lee et al., 2013).

Conclusion

The purpose of this study was to compare kinematic gait characteristics in FW and BW with and without the presence of CMI. While there was a significant difference across all analyzed gait characteristics based on the direction of movement, the cognitive task had no significant effect on them. FW showed longer stride length, more time in stride phase, and less time in swing phase than BW. There are not many studies that compare the effects of both CMI and directionality on kinematic gait characteristics across multiple conditions (FW, BW, FW with load, BW with load). An interesting finding in this study was that the stance and swing phase were significantly different across direction. Future studies can also implement kinetic analysis within these different experimental groups to help explain why kinematic gait characteristics were different as well as use a different task-presentation order to further explore the effects of task-prioritization on gait.

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