2023-09-22

Quantitative Assessment of Gait During Rehabilitation Using an Instrumented Treadmill

Fitzsimons, Karson

UNIVERSITY OF CALGARY

Quantitative Assessment of Gait During Rehabilitation Using an Instrumented Treadmill

by

Karson Fitzsimons

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE

DEGREE OF MASTER OF SCIENCE

GRADUATE PROGRAM IN BIOMEDICAL ENGINEERING

CALGARY, ALBERTA

SEPTEMBER, 2023

Abstract

Kinetic gait analysis of subacute stroke is a relatively unexplored area of study. Chronic stroke literature on the subject is extensive but does not capture the time period where the extent of recovery is greatest. Translating methods of gait analysis seen in research to a clinical setting is subject to many additional requirements which have previously prevented such investigations. The work presented in this thesis represents the first investigation using NeuroRecoVR, a new instrumented treadmill facility located within an inpatient rehabilitation gym. Working directly with inpatient physiotherapists, this study examines kinetic based gait parameters to quantify levels of impairment in subacute stroke. Recovery is most readily seen in changes in the walking speed of an individual, with many other gait parameters changing alongside walking speed. Therefor the relationships for all parameters of interest to walking speed are investigated in both neurologically intact controls (n = 14) and those undergoing rehabilitation for subacute stroke (n = 15). Parameters including spatiotemporal measures, forces, impulses, center of mass trajectory, center of pressure variability, and measures of symmetry were calculated for both groups. Subacute stroke participants have higher levels of asymmetry, increased instability, and altered gait dynamics compared to neurologically intact controls. The extent of recovery for each parameter was examined in a subset of stroke patients who took part in instrumented treadmill training over 1-2 months of rehabilitation (n = 4; mean ±SD age = 65 ±17; mean ±SD days post stroke at first session = 79 ±67). These participants showed improvements in stability, walking speed, and symmetry over the course of rehabilitation. These results show the benefit and potential for the use of kinetic analysis for aspects of both research and rehabilitation.

Preface

This thesis is original, unpublished, and independent work by the author, K. Fitzsimons. The experiments described in Chapters 3-5 were covered by Ethics Certificate #REB21-1576, issued by University of Calgary Conjoint Health Ethics Board for the project "Neurological recovery supported by virtual reality and gait training."

Acknowledgements

I would like to thank my supervisor Dr. John E.A. Bertram for his guidance and assistance in all aspects of this project. From the concept to the execution, nothing would have been possible without him. No matter how stressful the trials and tribulations of this research, a few words with him would always provide reassurance and clarity.

Thank you to Alexa Boyer and Dr. Deepthi Rajasekhar, for all aspects of recruitment and many long-shared days of data collection.

Thank you to Dr. Kelly Kaiser, Dr. Sean Dukelow, Dr. Gregor Kuntze, and many others from the RESTORE Network who made this research opportunity possible.

Thank you to all the PTs of unit 58 in the Foothills Hospital who worked so closely with us on the design of NeuroRecoVR, their assistance in data collection, and for providing invaluable feedback.

Thank you to Edith Rodie and her estate, whose remarkable gift made the RESTORE Network, NeuroRecoVR, and this research possible.

Table of Contents

List of Tables

List of Figures

List of Symbols, Acronyms, and Abbreviations

Symbol	Definition
AP	Anterior-Posterior
ML	Medial-Lateral
DS	Double Support
IC	Initial contact
FO	Foot-off
GRF	Ground Reaction Force
SD	Standard Deviation
KDE	Kernel Density Estimation
CoP	Centre of Pressure
CoM	Centre of Mass

Chapter 1. Introduction

1.1. Motivation

Quantitative gait analysis of subacute stroke (<6 months) is rarely seen due to the difficulty of translating many of the gait analysis methods commonly seen in research to a clinical setting. This has limited investigations into the longitudinal changes in certain gait parameters during the key period of recovery poststroke. The gold standard laboratory-based gait analysis techniques which combine kinetic and kinematic methods can be time consuming in both setup and post processing of the data. Kinematic analysis using marker-based motion capture requires time and training to accurately place markers on body segments of interest. This level of setup is not feasible in the timeframe available for inpatient rehabilitation, which is usually around 30 minutes, as it would take away too much time from the walking therapy. This setup time can be reduced by using markerless motion capture, though current systems have slightly reduced accuracy compared to marker-based methods (Kanko, et al. 2021). Situations common for inpatient settings including loose clothing and having multiple individuals in view can impact the accuracy of data from markerless systems (Wade, et al. 2022).

This leaves kinetic analysis as a potential method for quantifying gait variables within the time and facility constraints required. Force plate(s) require repeated passes to generate viable amounts of information and require cognisant foot placement on the force plate(s) in question. The alternative for this is to use an instrumented treadmill, allowing for continuous walking alongside data collection. Using an instrumented treadmill, we can capture the interaction between a person and the environment that they are walking on, giving us key information regarding the body's response to changes in gait. This requires less set up time than kinematic methods, and clinicians can work directly with the patients while data collection is ongoing. Using NeuroRecoVR, an instrumented treadmill facility in the rehabilitation gym of the Foothills Medical Center in Calgary, Alberta, this thesis will explore

the gait parameters which can be quantified from kinetic data alone in inpatient stroke rehabilitation.

1.2. Thesis objectives

The hypothesis of this thesis is that kinetic analysis of subacute stroke gait will provide insights into the impact of gait deficits on the dynamics of locomotion poststroke. Metrics derived from kinetic data will allow for gait to be characterized in a low number of steps and for changes in the metrics to be tracked across multiple sessions.

The main objectives of this thesis are:

1) Examine the metrics that can be calculated solely from the outputs of an instrumented treadmill and establishing which have speed-based relationships in neurologically intact populations at the slow speeds seen in subacute stroke populations (0.23-0.55m/s).

2) Compare the speed relationships seen in the neurologically intact populations to those living with subacute stroke.

3) Explore the changes in these gait metrics over the course of inpatient stroke rehabilitation.

1.3. Thesis structure

Chapter 2 provides a literature review of common methods of gait analysis in research, including spatiotemporal measures, kinetics, and kinematics. Characterization of the center of mass and center of pressure trajectories are discussed. The impact of stroke on human walking and how it affects the gait parameters found in research of chronic stroke is examined.

Chapter 3 examines the impact of slow walking on gait parameters in neurologically intact individuals. This includes discussion on which parameters are correlated with changes in walking speed, which will be used for comparison in chapter 4.

Chapter 4 examines the same parameters discussed in chapter 3 applied to stroke patients undergoing inpatient rehabilitation. General trends seen across poststroke participant are highlighted. Longitudinal changes to gait parameters in a subset of poststroke participants are examined in detail.

Chapter 5 discusses the implications of instrumented treadmill-based gait analysis for stroke rehabilitation. Limitations and suggestions for improvements to the system are discussed.

1.4. Statement of contribution

The data collection for this thesis was completed as part of a larger pilot study on the impact of split-belt treadmill training in stroke rehabilitation. Alexa Boyer, Deepthi Rajasekhar, and the author shared responsibility for data collection across all participants.

All code used for data analysis is an original and independent work by the author, with the exception of the kernel density estimation function from (Botev, Grotowski, & Kroese, 2010).

Chapter 2. Literature review

2.1. Neurologically intact gait

Human locomotion is an often-underappreciated aspect of life that can appear outwardly simple, whereby one moves their legs repeatedly in such a way that causes them to move forwards. The mechanisms behind moving one's legs introduce new complexity, what muscles to activate, when to activate them, the forces they exert, and the stability of the motion (Kuo & Donelan, 2010). No matter the complexity of the mechanisms and the relationship behind them, the goal of locomotion remains the same (Tesio, Rota, & Perucca, 2011). In the case of walking, this goal can be represented as managing the trajectory of the body's centre of mass (CoM) in the most efficient stable manner given the circumstances (Croft, Schroeder, & Bertram, 2017).

This goal helps explain why given a near infinite combination of parameters for walking, humans will often default to a stereotypical gait for a given circumstance. Constraining a given parameter such as walking speed or frequency causes an individual to adjust the other parameters to fit an optimal solution energetically (Bertram, 2005). The idea of human locomotion being energetically optimized has been explored in both modeling and experimental studies. It also provides context for certain aspects of gait such as pre-emptive push off as it relates to reducing collisional losses in the step-to-step transition (Donelan, Kram, & Kuo, 2002).

2.1.1. The gait cycle

Human movement is often defined in terms of phases in the gait cycle over a stride. For the context of this thesis, a stride is defined as the period between two consecutive initial contacts of the same foot. The gait cycle is assumed to be periodic in steady state walking. The initial contact (IC) and foot-off (FO) of each step are key events during the gait cycle that are used to segment it into different phases. The identification of these gait events can be done using kinematics such as maximum positions or when velocities of specific markers

switch directions (Fellin, Rose, Royer, & Davis, 2010), but the use of ground reaction force is considered the gold standard (Zeni, Richards, & Higginson, 2008) (Rouhani, Abe, Nakazawa, Popovic, & Masani, 2015). When the foot is in contact with the ground it is considered to be in stance phase, as opposed to the swing phase when it is off the ground. When both feet are in contact with the ground it is termed double stance or double support.

2.1.2. Role of upper limb

In neurologically intact gait, there is a natural tendency for the upper limbs to swing contralateral to the leading leg. There are contending theories of the primary role of the upper limbs related to stability and energetic cost. It has been shown that the swing of upper limbs reduces the vertical moment, the torque which causes the body to twist about a vertical axis, which decreases the energy expended while walking (Collins, et al., 2009). Alternate theories include upper limb swing is used to increase stability in the lateral direction or to improve the ability to recover after a perturbation (Bruijn, et al., 2010).

When walking at lower speeds, the energetic benefit of swinging the upper limb is diminished (Thomas, et al., 2020). The impact of upper limb swing on perturbation recovery and stability was found to be maintained at reduced speeds (Bruijn, et. al. 2020).

2.1.3. Spatiotemporal parameters

Spatiotemporal parameters of gait include comprehensive measures relating to both sides and unilateral measures relating to either side of the body. Comprehensive measures include walking speed, stride length, cadence, and double support time. Unilateral measures include step lengths, step times, single support, stance, and swing times. Variability in the frontal plane, specifically step placement examined via step width variability, has been used to establish the requirement of active balance control in the lateral direction (Bauby & Kuo, 2000). Step length, cadence, and walking speed form a relationship where if any two are known the third can also be determined. Cadence or step frequency is constrained when

walking, to the beat of a metronome or music, step length can be constrained via markers on the floor, and walking speed is constrained when on a treadmill.

2.1.4. Kinematic derived parameters

In the field of gait analysis, kinematics is often associated with motion capture describing the joint angles, positions, accelerations, and velocities of the various segments of the body. This allows for within limb analysis of the mechanisms behind locomotion. Motion capture often requires markers to be placed on anatomical landmarks of each body segment to accurately gauge the relative motion using multiple cameras placed around the space. These markers can introduce error when placed incorrectly or if they move on the skin during data collection. Recent innovations in markerless motion capture have shown promise in approaching the accuracy of marker-based methods (Nakano, et al., 2020). Though currently not implemented in this study, future advancements in markerless motion capture could prove beneficial to the system used in this thesis.

2.1.5. Kinetic-derived parameters

Kinetics describes the relationship between motion of a body and the forces which cause it. The motions induced by the forces and how they vary describe the dynamics of the system. In gait analysis this is traditionally measured using force plates which can measure the interaction between an individual and the environment via the forces/moments they exert while walking. Compared to the within limb analysis of kinematics, kinetics allows for the examination of the task locomotion is looking to achieve, managing the CoM trajectory. The ground reaction forces (GRFs) of human walking have a stereotypical shape in each direction. Shown in Figure 2-1 are an example of the vertical and anterior-posterior (AP) GRFs. The vertical GRF for one leg while walking is characterized by the double peak shape.

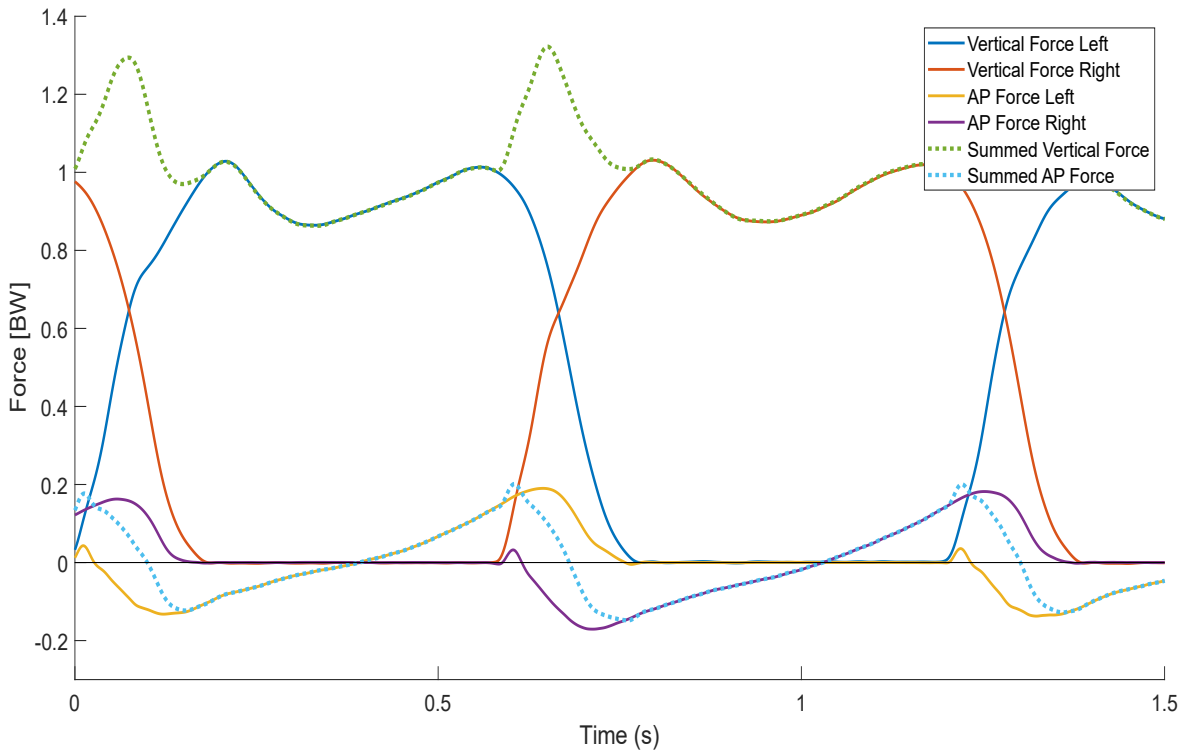

Figure 2-1: Vertical and anterior-posterior (AP) ground reaction forces for individual legs while walking and the sum of forces in each direction.
Recorded from a single neurologically intact individual at 1.0m/s walking speed. Forces are normalized to body-weight of the individual.

The AP GRF consists of a negative portion corresponding to braking and a positive portion corresponding to propulsion. The medial-lateral (ML) GRFs are generally of smaller magnitude and are often directed laterally on each side. The summed forces from both leg in each direction represent how an individual is interacting with the environment and therefore the resulting forces acting on their CoM (Donelan, Kram, & Kuo, 2002).

There are several useful features of the GRF curve that can be used during gait analysis. Firstly, the forces can be integrated over time to determine the impulse applied in each direction (Goldberg, Kautz, & Neptune, 2008). For example, in steady state walking the vertical impulse over the gait cycle (area under the summed vertical force trace) is equivalent to the impulse exerted by the individuals body-weight. The area under the AP GRF curve is equal to the propulsive (positive) and braking (negative) impulse.

In this thesis, a split-belt instrumented treadmill (Bertec Corporation, Columbus, Ohio) which contains a force plate under each belt is used as the walking surface for all ground reaction forces. The validity of treadmill gait to describe overground walking has been investigated, and slight differences have been found between the two (Sloot, Houdijk, & Harlaar, 2015) (Lee & Hidler, 2008). This can be reduced by using a self-paced treadmill where walking speed dynamically adjusts to the user (Sloot, van der Krogt, & Harlaar, 2014). Benefits for engagement in treadmill training have been seen when VR environments are integrated in the walking, and participants rated the experience as closer to overground walking (Sloot, van der Krogt, & Harlaar, 2014).

2.1.6. CoM trajectory metrics

In kinematic studies, the motion of the body's CoM is often approximated with a sacral marker or weighted averages of body segment motion (Gard, Miff, & Kuo, 2004). Alternatively, the total GRF in each direction can be used to determine the motion of the CoM, characterizing the previously stated goal of locomotion using only kinetic data. The sum of forces in each direction divided by the body mass will give the resulting acceleration of the CoM (Cavagna, 1985). Integrating these accelerations over time will provide the velocity of the CoM (Donelan, Kram, & Kuo, 2002). Integration constants for the ML velocity (Eq 1.1) and vertical velocity (Eq 1.3) are determined by making the average velocity in those directions equal to zero. This is done under the assumption that the average displacement in each of those directions over a stride is zero. The AP direction constant is determined by making the mean forward velocity over the stride equal to the walking speed. This method determining of CoM trajectory has been shown to be more accurate compared to kinematic methods (Whittle, 1997) (Gard, Miff, & Kuo, 2004).

$$v_x = \int a_x \, dt = \int \frac{F_{xL} + F_{xR}}{m} dt \qquad (1.1)$$

$$v_y = \int a_y \, dt = \int \frac{F_{yL} + F_{yR}}{m} dt \qquad (1.2)$$

$$v_z = \int a_z \, dt = \int \frac{F_{zL} + F_{zR} - mg}{m} dt \qquad (1.3)$$

The instantaneous mechanical power generated by the individual limbs can then be calculated by the dot product of the CoM velocity vector and the ground reaction force vector (Donelan, Kram, & Kuo, 2002). Further integration of the mechanical powers generated by each limb will result in the mechanical work.

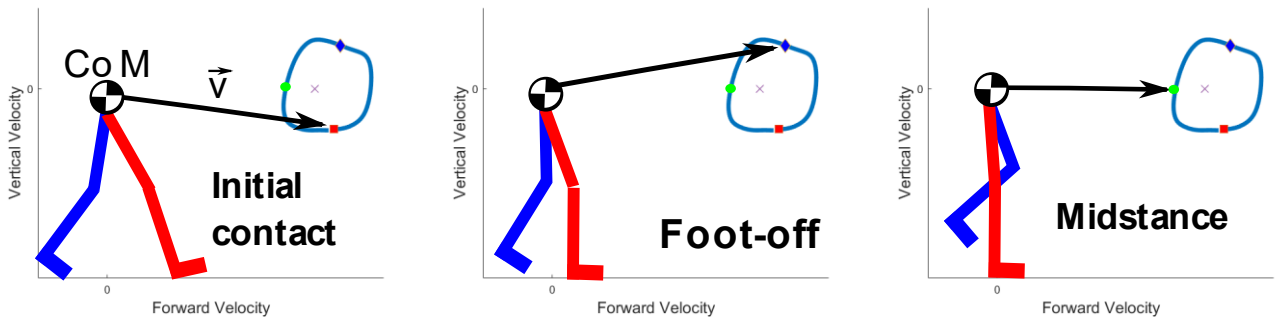

Figure 2-2: Saggital plane CoM velocity hodograph. The vector tip follows a counterclockwise loop for walking at average speeds (~1 m/s).

The tip of the velocity vector of the CoM over the course of the gait cycle can be visualized using a hodograph (Adamczyk & Kuo, 2009) as shown in Figure 2-2. The hodograph has been used in previous studies to analyze the step-to-step transition in young neurologically intact individuals, amputees (Adamcyzk, 2008), and elderly adults (Meurisse, Bastien, & Schepens, 2019a).

The step-to-step transition is defined as the period between when vertical velocity reaches its minimum at or before initial contact and maximum at or after foot off (Adamcyzk, 2008). This period corresponds to when the CoM velocity is being redirected while negotiating the transition onto the leading leg for the following step. As an individual moves through single stance, they fall forward and downward onto the next stance limb, shown as a downward angle on the hodograph when vertical velocity is negative. Approaching initial contact, the CoM trajectory no longer accelerates downward as pre-emptive push-off is initiated to redirect the velocity, this is accompanied by an increase in forward velocity.

Following initial contact, as the bodyweight shifts onto the following stance leg, the CoM trajectory is redirected upward to a positive vertical velocity. The CoM accelerates upward during double support and allows for the pendular motion during stance after the trailing foot-off.

The “D” shape of the hodograph in the sagittal plane for walking follows the tip of the CoM velocity vector in a counterclockwise loop. As speeds approach and pass the walk-run transition the sagittal hodograph will invert itself into a figure-eight before becoming a clockwise loop. This has been demonstrated in a study of avian gait (Usherwood, 2010). Quantitative characterizations of the hodograph include the angle that velocity is redirected in the step-to-step transition, timing of the step-to-step transition relative to initial contact, the acceleration, and the velocity at specific time points in the gait cycle (Soo & Donelan, 2012). Corresponding to the integration constants used when determining the CoM velocities, the mean velocity for the forward direction is the walking speed while the vertical and lateral direction have means of zero velocity.

At slower speeds the step-to-step transition seen in the hodographs has been shown to match the double support period. Meaning that the CoM velocity redirection begins at or after initial contact (Meurisse, Bastien, & Schepens, 2019). This corresponds to a reduced use of pre-emptive push-off at slow speeds.

2.1.7. Centre of pressure

The centre of pressure (CoP) indicates the point of application for the resultant GRF vector on the force plate. The trajectory of the CoP in standing balance has been used as an assessment of postural control termed the stabilogram. In walking, the CoP trace on the force plates of an instrumented treadmill can be plotted as a cyclogram or butterfly diagram (Kalron & Frid, 2015).

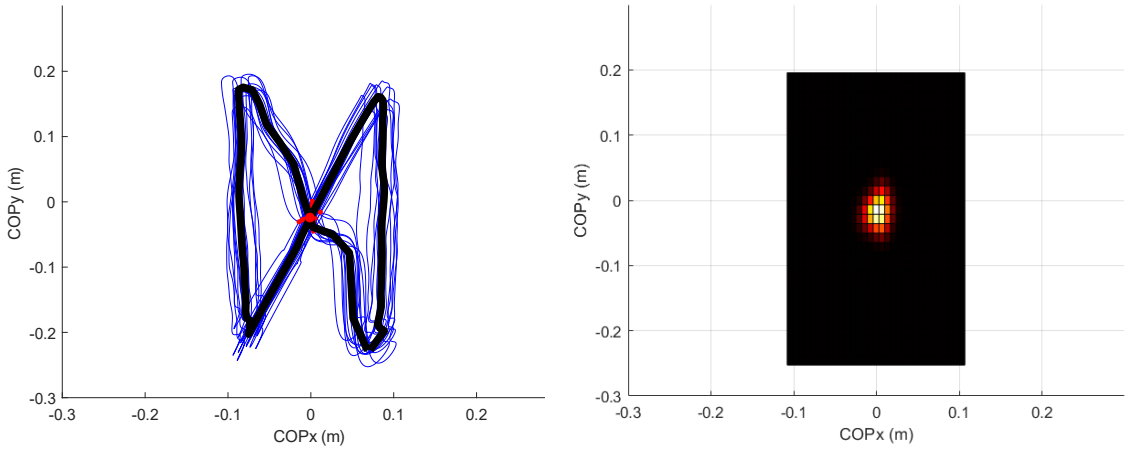

Figure 2-3: Butterfly diagram and intersection density plot for a single neurlogically intact individual

Left: The butterfly diagram shows traces of the CoP over multiple strides (blue) and the mean of these traces (black). The location of intersections of the diagram are marked with red circles. Right: The 2D kernel density estimation plot for the butterfly diagram. It quantifies the density (red to white meaning higher) of intersections and the range of CoP values (black box).

The butterfly diagram, as seen in Figure 2-3, has been used to represent multiple aspects of gait with a single plot. These can include gait variability, stride width, single/double support, and weight shift symmetry (Kalron & Frid, 2015). The butterfly diagram features an intersection in the CoP displacement during the double support periods of consecutive steps. The variability of the intersections of the butterfly diagram is often used to describe gait variability in the AP and ML directions. Another method of characterizing the variability of these intersections is 2D kernel density estimation (Lee & Liang, 2020). This provides a quantitative measure of the density of intersections within the butterfly diagram and helps visualize the variability of this density in both directions combined.

2.1.8. Analysis of slow walking

Self-selected walking speed is known to decrease with age, injury, and neurological impairments (Meurisse, Bastien, & Schepens, 2019). Most kinematic and kinetic-based gait research on human walking is conducted above 0.8m/s. This range covers the average walking speed of a neurologically intact individual (1.2-1.4m/s) and extends up until the walk-run transition near 2.0m/s. Available datasets containing kinematic and kinetic measures often set their slow gait speed around 0.8m/s (Moore, Hnat, & van den Bogert, 2015). One study on the stability of slow walking (Bruijn, van Dieën, Meijer, & Beek, 2009), with a slowest speed of 0.62m/s, found less stable gait at slower speeds but was unable to draw significant relationships.

More recent studies have sought to examine the impact of slow walking to fulfil the need for normative kinematic and kinetic data at these lower ranges. Smith (2018) recorded kinematics and kinetics for thirty young neurologically intact participants at walking speeds between 0.2 and 0.8m/s. Their analysis was focused on joint level angles and power. They found that there were inconsistent inflection points within the parameters that indicated individuals may adapt to slow speeds differently. The results from Smith (2018) were more tailored to the use of lower extremity powered exoskeletons for spinal cord injury, establishing low speed regressions for joint trajectories (Smith, Lemaire, & Nantel, 2018).

Wu (2019) provides a dataset for eight neurologically intact young individuals between 0.1 and 0.6m/s. This study reports common spatiotemporal measures such as step length, step width, and double support period. The study also examines CoM work rate using the methods described in Donelan (2002) that are discussed in section 2.1.6. The results of the study focus on the sagittal plane impacts of slow walking speed and found that most speed related trends in that plane can extend to slower speeds. The exception to this was step width variability and the CoM work during the preload phase prior to push-off. A follow-up using the same dataset (Best & Wu, 2020) investigated the lateral stability impacts of slow walking. This study found that the lateral range of CoM velocity tended to remain constant while measures of stability in single stance decreased at lower speeds. They theorized that in the

young neurologically intact individuals studied, a lower margin of stability was being adopted instead of adopting a limited lateral CoM velocity due to increases in the energetic cost.

2.1.9. Nondimensionalization of gait metrics

In order to accommodate for differences in body size when comparing gait metrics between individuals, it is common to normalize gait metrics by certain body proportions. With respect to treadmill walking, nondimensional walking speed (v_{ND}) is shown in equation 1.4, where the treadmill speed (v) is divided by the square root of gravitational acceleration (g) multiplied by leg length (L). All other gait metrics can be nondimensionalized using body mass, gravity, and leg length as base units.

$$v_{ND} = \frac{v}{\sqrt{gL}} \qquad (1.4)$$

2.2. Gait poststroke

2.2.1. Prevalence and types of stroke

Stroke is an emergency medical condition that occurs when the blood supply to portions of the brain in interrupted. This interruption to the supply of oxygen and nutrients to the brain can lead to brain damage, disability, and death. There are two main categories of stroke, ischemic and hemorrhagic. Ischemic stroke occurs when the blood supply to an area of the brain is cut off, whereas hemorrhagic is caused by the rupture of a blood vessel. Ischemic stroke is more common, representing approximately 80% of cases (Donkor, 2018).

Stroke is the second-leading cause of death worldwide and results in up to half of survivors experiencing long-term disability (Donkor, 2018). The most common impairments that occur following stroke are difficulties in communication (aphasia), unilateral weakness (hemiparesis) or paralysis (hemiplegia). The latter two are responsible for the 75% of

poststroke individuals who experience long-term limitations to mobility (Sheffler & Chae, 2015). This reduced mobility is often associated with lower quality of life and poorer functional independence (Rowland & Lewek, 2022). The impact of stroke is often complicated by the higher occurrence in older individuals. Gait and balance function are known to decline with age (Osoba, Rao, Agrawal, & Lalwani, 2019), but this decline may be exacerbated by even minor stroke events.

Following a stroke, the stages of recovery are divided into three phases. The first week poststroke is termed the acute phase, where many of the imminent dangers of a stroke are still present. Between one week and 6 months is the subacute phase, when much of inpatient physical and occupational rehabilitation occurs. After 6 months is the chronic phase which is primarily impacted by the long-term deficits of stroke.

2.2.2. Impacts on gait poststroke

The cause of gait impairments has been theorized to be combined effect of deficits in neural control, loss of proprioception, ataxia, and muscle weakness (Sheffler & Chae, 2015). Impairments related to sensory interpretation, neuromuscular control, and attention will also effect gait. The impact of limited range of motion due to spasticity or muscle weakness is most readily apparent at the ankle joint, where foot drop and propulsion issues are common poststroke gait deficits.

The severity of stroke and its resulting impairments are often related to the type and location of the stroke (Jones, et al., 2016). An example of this is shown in split-belt treadmill walking studies, where those who suffer cerebellar strokes are unable to adapt to the split-belt task (Finley, Long, Bastian, & Torres-Oviedo, 2015). Injury or dysfunction in the cerebellum is known to have impacts on motor learning, adaptation, balance, and coordination (Thach & Bastian, 2004).

Neural control of gait combines multiple sources of sensory information to influence both automatic and conscious aspects of gait (Takakusaki, 2017). Portions of the brainstem related to gait include central pattern generators which are theorized to affect steady state

aspects of locomotion (Guertin & Steuer, 2009). Sensory information from the parietal, occipital, and temporal lobe are combined with prior experiences and knowledge of the body to allow relevant neuromotor signals from the motor cortex, brainstem, and cerebellum. Damage to any of these portions of the brain, and other associated areas will cause deficits to gait (Takakusaki, 2017).

Deviations from neurologically intact gait patterns in stroke are known to cause increased metabolic cost for walking (Penke, Scott, Sinskey, & Lewek, 2019). This includes impairments to propulsion and gait symmetry affecting the overall efficiency, leading to reduced endurance poststroke. This can be related to increased losses due to impaired step-to-step transition dynamics and reduced pre-emptive push-off. Other studies make the distinction that hemiparetic gait is not less efficient but has greater metabolic cost (Farris, Hampton, Lewek, & Sawicki, 2015), prompting debate on how to define gait efficiency.

2.2.3. Gait rehabilitation poststroke

The impact of stroke on gait is often immediate and can have long-term detrimental effects. The Copenhagen Stroke Study by Jorgensen and colleagues (1995) tracked walking ability from hospital admission to rehabilitation discharge. They found that at hospital admission, around 51% of patients had no walking function and that that only 65% of individuals were able to achieve independent walking ability after rehabilitation. The outcome and speed of recovery in this study was highly dependent on the initial level of gait impairment.

The timing and intensity of rehabilitation is a key factor in the recovery of lost walking ability (Teasell, Bitensky, Salter, & Bayona, 2005). The major recovery of walking ability, if any, is seen in the first 11 weeks poststroke (Jorgensen, Nakayama, Raaschou, Olsen, & Jergensen, 1995). Functional gains often continue within the first 6 months before tapering off as the stroke transitions into the chronic phase (Alexander, et al., 2009). The initial 10-week period following a stroke is considered to be the peak of neurological recovery and often coincides with inpatient rehabilitation. Delayed access to physiotherapy

has been found to reduce the level of motor recovery observed, with hemorrhagic stroke patients showing no improvements after therapy sessions during the chronic phase (Callegari, et al., 2021). A recent study has shown that higher intensity walking during inpatient rehabilitation improves both walking and quality of life outcomes during the first year poststroke (Klassen, et al., 2020). Achieving independent walking ability is often listed as a primary goal for inpatient rehabilitation and remains a high priority in the chronic phase (Harris, Eng, & Ot, 2004).

2.2.4. Clinical outcome measures

In clinical settings there exist many assessments that have been created to describe functional outcomes poststroke. Often used are observation based ordinal scales where patients are required to complete a series of tasks or movements. These include the Barthel Index, Fugl-Meyer Assessment Scale (Fugl-Meyer, Jaasko, Leyman, Olsson, & Steglind, 1975), Chedoke-McMaster Stroke Assessment (Gowland, et al., 1993), Functional Independence Measure, Berg Balance Scale, and many others (Mohan, et al., 2021). The number and type of scales in use will often vary with treatment center, but they are often very prevalent due to the low cost and short time to administer the assessment. These scales can often suffer from ceiling effects. In addition, due to the nature of ordinal scales, the difference between measurement values does not reflect equivalent measures of functional improvement (Hawe, Scott, & Dukelow, 2019).

Regarding walking function, there are other common assessments used alongside the ordinal scales. The 10-metre walk test (10MWT) quantifies walking speed as the time it takes to walk a 10m distance with an extra 2m given for initial acceleration, and the 6-minute walk test (6MWT) quantifies walking endurance of a patient by how many meters they are able to walk in 6 minutes (Teasell et al., 2020).

2.2.5. Spatiotemporal changes poststroke

The most apparent change in gait poststroke is the reduction in walking speed. Average self-selected walking speed for individuals poststroke is in the range of 0.23-0.95m/s depending on level of impairment (Beyaert, Vasa, & Frykberg, 2015). A common benchmark for stroke rehabilitation is 0.8m/s walking speed. which is the requirement for community ambulation, often cited as the speed needed to cross a street in time (Teasell et al., 2020). The relationship for cadence and step length has also been shown to be altered poststroke, with poststroke individuals selecting greater than or equal to cadences than neurologically intact individuals for the same walking speeds. The stride length correspondingly sees a decrease at these higher cadences. Because many parameters of gait scale with walking speed, it is often difficult to determine which aspects quantify recovery (Wonsetler & Bowden, 2017).

Gait parameters in stroke research often introduce aspects of symmetry in addition to raw measurements from each leg. Most often, symmetries are reported as the proportion of the paretic side's contribution compared to the sum or as the ratio of paretic to non paretic contributions (Mahon, Farris, Sawicki, & Lewek, 2015). These measures of symmetry are theorized to be more sensitive and able to reflect the changes in stroke recovery compared to solely reporting the measures from a single limb. The variety of symmetry measurements used in the literature makes it difficult to compare between studies, and the effect of asymmetry is not yet thoroughly understood. The direction of step length asymmetry, whether the paretic takes larger or smaller steps, has been correlated with plantarflexor impairment and compensatory gait strategies (Allen, Kautz, & Neptune, 2011).

2.2.6. Kinetic measures poststroke

The kinetic measures of gait in chronic stroke research often focus on sagittal plane interactions. Both propulsive forces and impulses have been used to assess chronic stroke (Bowden, Balasubramanian, Neptune, & Kautz, 2006). The propulsive force and impulse of the affected side has been found to decrease poststroke while the braking component

increases (Lewek, Raiti, & Doty, 2018). This decrease in propulsion is highly linked to walking speed and the duration of swing phase of the paretic leg (Dean, Bowden, Kelly, & Kautz, 2020). The decrease in propulsion is quantified in post-chronic stroke by either the peak propulsive force or the propulsive impulse and the ratio of paretic to total propulsion. Some studies link cadence to propulsive impulse in relation to walking speed (Hsiao, Zabielski, Palmer, Higginson, & Binder-Macleod, 2016) with improved linear fits. CoM work is often calculated alongside propulsion in chronic stroke studies, with reduction in both work and propulsion reported for the paretic leg (Mahon, Farris, Sawicki, & Lewek, 2015).

2.2.7. Poststroke recovery

Gait impairments often lead to adaptations in human gait. In the context of stroke, these are divided between functional and neurological recovery (Beyaert, Vasa, & Frykberg, 2015). Functional recovery involves the use of compensation methods including hip-hiking, circumduction, and vaulting, all of which are examples related to improving foot clearance with an altered movement strategy.

Neurological recovery, also called spontaneous recovery, is related to the areas of the brain directly damaged by a stroke. As blood supply is restored or swelling (edema) subsides, some tissues are able to resume function if they are not irreversibly damaged, this process occurs during the initial recovery period poststroke (Teasell et al., 2020). Recovery continues after this process through reorganization of the connections within the central nervous system to accommodate for the damaged tissue, termed neuroplasticity.

2.3. NeuroRecoVR system description

Figure 2-4: NeuroRecoVR treadmill

The NeuroRecoVR treadmill facility, pictured in Figure 2-4, combines a Bertec split-belt instrumented treadmill with semi-immersive VR. The screens (Samsung, Suwon, South Korea) display custom designed modules which are designed to gamify rehabilitation (Red Iron Labs, Calgary, AB). The system includes an overhead frame which holds a two-axis body-weight support harness, and wheelchair lift for patients to reach the elevated walking surface. All data collection for this thesis was completed using the same VR module, an animated walking path, with no biofeedback shown to avoid influencing the data.

Chapter 3. Kinetic analysis of slow walking gait in neurologically intact individuals

3.1. Methods

3.1.1. Participants

Fourteen individuals (9 female, 5 male), with no history of neurological impairment and no recent leg injuries were recruited. Recruitment was completed via advertisements at local community organizations and word of mouth. Participants were required to be age 18 or older and able to give informed consent. Participants were excluded from the study if they had recent lower limb injury, previous neurological injury, gait impairments, and other conditions which would prevent safe participation or impact their ability to walk on the treadmill.

The mean (±SD) age of participants was 34 (±17) years old, mean (±SD) body mass was 76 (±18) kg, with a mean (±SD) leg length of 88.0 (±5.0) cm measured from greater trochanter to medial malleolus averaged across both legs. The mean (±SD) self-selected walking speed of participants was 1.38 (±0.22) m/s (0.47 ± 0.1 Non dimensional) assessed via an overground 10MWT. All other parameters were nondimensionalized using body mass, gravity, and leg length as base unless otherwise noted.

3.1.2. Protocol

After comfortable walking speed was assessed, participants walked on a split-belt instrumented treadmill (Bertec Corp, Columbus, Ohio) at 4 different speeds in a randomized order. The randomized order was determined via an online tool which utilized a pseudo-random number generator (Urbaniak & Plous, 2013). The speeds were 0.23m/s, 0.39m/s, 0.55m/s, and their self-selected speed adjusted for comfortability on the treadmill. The three low speeds represent the mean ± one standard deviation for walking speed in subacute stroke rehabilitation. Participants walked for two minutes at each speed while a semi-immersive virtual reality pathway was displayed on screens in front of them. After the tied-belt walking

was completed, participants completed a split-belt walking protocol at two of the slow speeds as part of another study.

3.1.3. Data collection and analysis

GRF data was sampled at a rate of 1000 Hz from each belt. All processing and analysis was completed using MATLAB (MathWorks, 2021). The GRF data was low pass filtered at 25 Hz with a zero-lag 4th order Butterworth filter. Gait events including initial contact and foot-off of each leg were identified based on a 20N threshold across a 40-sample window. Any steps which contained crossovers onto the opposite belt or did not follow a normal order of IC and FO were flagged. The longest consecutive streak of uninterrupted strides was then determined, and the 15 consecutive strides of this streak were used for analysis. 15 strides were used as the range for analysis as it represented the minimum number of strides required in the analysis of stroke gait presented in chapter 4 and has been cited as an acceptable number for the calculation of gait variability (Kribus-Shmiel, Zeilig, Sokolovski, & Plotnik, 2018) (Kroneberg, et al., 2019).

Step length was taken as the treadmill velocity multiplied by the time between contralateral initial contacts. Step width was measured as the lateral distance between the CoP at contralateral initial contacts. Double support time was measured as the proportion of the stride where both feet are in contact with the treadmill. Propulsive peak and impulse were taken from the AP direction GRF. Lateral impulses were determined by the ML direction GRF. Maximum kernel density estimate and CoP intersection variability were calculated using the methods described in section 2.1.7 (Kalron & Frid, 2015) (Lee & Liang, 2020). CoM acceleration and velocity were determined using the method outlined by Cavagna (1975) and Donelan (2002). The standard deviation of each metric over the 15 strides was also calculated as a measure of variability. Symmetry or absolute difference between left and right side for each metric and the standard deviations of the differences/symmetry were calculated as well for comparison to the stroke data in chapter 4. Measures of symmetry for neurologically intact individuals were taken to be the smaller value over the total of both sides to reflect the equivalence to paretic over total in stroke.

There are several gait parameters with known relationships to walking speed. Step length, cadence, and double support time have known nonlinear relationships with walking speed and were fitted to power law relationships to reflect this (Wu, Simpson, van Asseldonk, van der Kooij, & Ijspeert, 2019). Fits were performed across all calculated metrics and relevant relationships to walking speed or the lack thereof for closely linked metrics are presented in the Table 3-1-Table 3-4. For these tables all regressions were completed against nondimensionalized walking speed. The level of significance used for the fits is adjusted for multiple comparisons using a Bonferroni correction so that $\alpha = 0.002$. Significance levels for power law fits were not calculated, but power law fits were only used when confirmed in literature.

3.2. Results

3.2.1. Walking speed relationships

The fits for spatiotemporal measures of gait in Table 3-1 show that the expected nonlinear relationships for step length, double support time, and cadence fit the data well (R^2 >0.85). Step width variability shows significant relationships to walking speed, but a poor fit to the data ($R^2 = 0.195$). Step width did not show a significant relationship to walking speed.

Table 3-1. Regression results for spatiotemporal measures with respect to walking speed in neurologically intact participants

Metric	Fit Type	Coefficients (±95%CI)	Offset	R^2	Linear fit p-values
Step Length	Power ax^b	a: 1.208 ± 0.055 b: 0.525 ± 0.0313		0.906	
Step Width	Linear	-0.021 ± 0.064	0.242 ± 0.016	0.007	0.527
Step Width SD	Linear	0.013 ± 0.007	0.012 ± 0.001	0.195	0.0006*
Double Support Time (% stride)	Power ax^b	a: 0.179 ± 0.017 b: -0.448 ± 0.044		0.903	
Cadence (Hz)	Power ax^b	a: 2.765 ± 0.209 b: 0.463 ± 0.050		0.865	

Standard deviation (SD), cadence in Hertz (Hz) is steps per second, *significant result for linear fit

The plot of the regressions for the spatiotemporal regressions excluding cadence are shown in Figure 3-1. The three bands of lower speeds represent the trials at 0.23, 0.39, and 0.55 m/s, while the higher speeds (> 0.2 nondimensional speed) are the self-selected speeds of the individuals. The general relationships for spatiotemporal variables are as expected, with step length and cadence increasing with walking speed while double support time decreases.

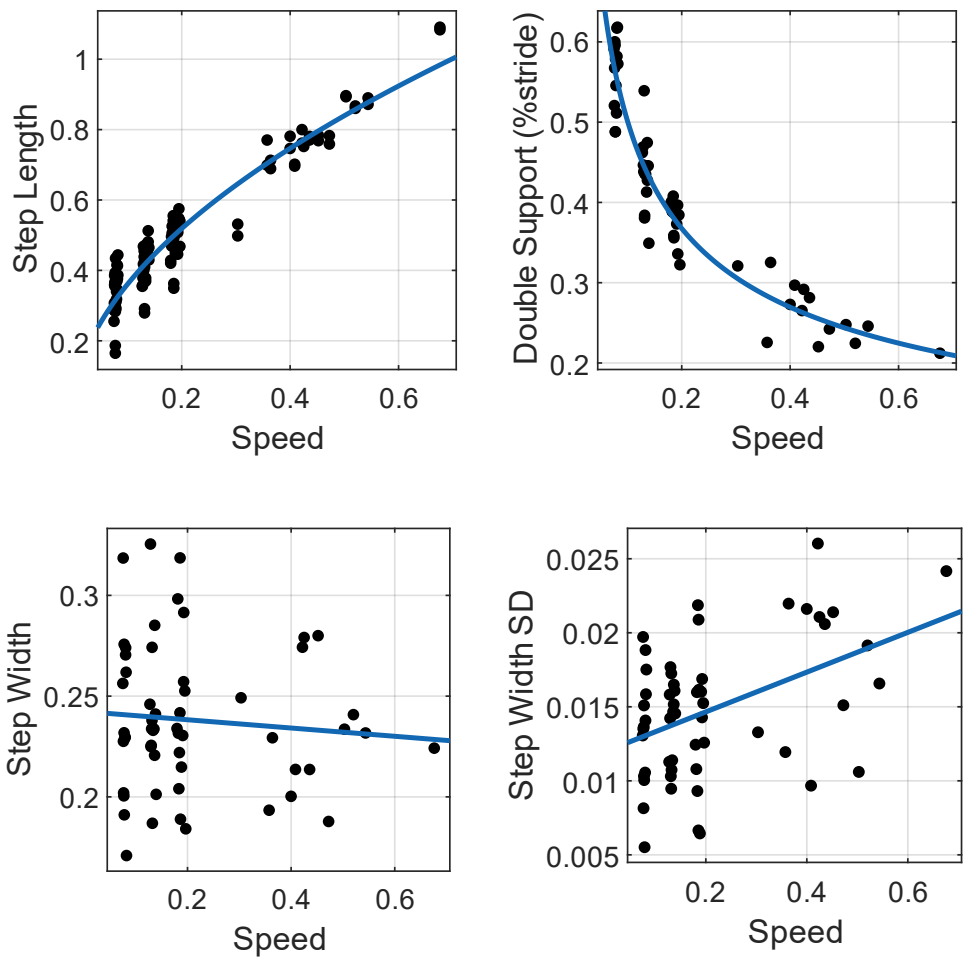

Figure 3-1: Regressions for dimensionless spatiotemporal gait measures with respect to dimensionless walking speed in neurologically intact participants

The walking speed regressions for forces and impulses is shown in Table 3-2. Significant linear relationships with good fits are seen for peak propulsive force, propulsive

impulse multiplied by cadence, and the mean vertical force during double support. Propulsive impulse on its own had less variance explained by walking speed (R^2: 0.291) but still increased linearly with walking speed. Lateral impulse showed a decrease with speed with a weaker fit to the data, this indicates that there is potentially increased active stability in the lateral direction at the slower walking speeds.

Table 3-2: Regression results for force and impulse measures with respect to walking speed in neurologically intact participants

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	p-value
Propulsive Impulse (BW*s)	Linear	0.024 ± 0.007	0.0	0.291	8.74e-10*
Lateral Impulse (BW*s)	Linear	-0.051 ± 0.012	0.0572 ± 0.003	0.375	7.12e-13*
Peak Propulsive Force (BW)	Linear	0.462 ± 0.021	0.005 ± 0.006	0.944	1e-70*
Propulsive Impulse x Cadence (BW)	Linear	0.115 ± 0.011	0.008 ± 0.002	0.900	1.24e-28*
Mean Vertical Force in double stance (BW)	Linear	0.643 ± 0.052	0.940 ± 0.013	0.846	1.6e-48*

Data was normalized with respect body-weight (BW) of individuals before regression, *significant result for linear fit

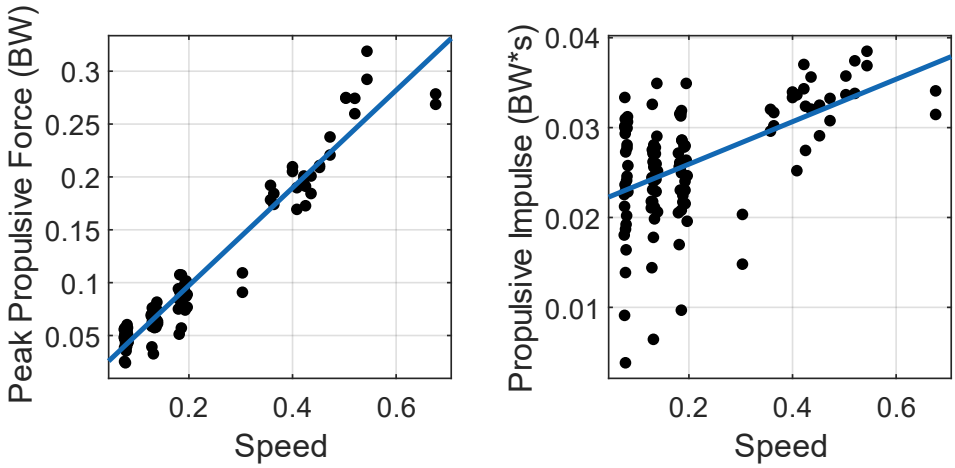

Figure 3-2: Regressions for propulsive force and impulse with respect to walking speed in neurologically intact participants

The two strongest relationships to walking speed among CoM trajectory metrics were the change in vertical velocity over the course of double stance and the acceleration of the CoM at initial contact. The vertical velocity difference quantifies how much the CoM velocity vector must be redirected, only half of the variance in the data is explained by walking speed ($R^2 = 0.564$). CoM acceleration showed a nonlinear relationship with walking speed, the quality of the fit is similar to the velocity difference ($R^2 = 0.556$). At the slow walking speeds, many of the participants had negative acceleration values at initial contact. This indicates that at slow speeds there is not much less pre-emptive push off compared to normal walking speeds.

Table 3-3: Regression results for select CoM trajectory metrics to walking speed in neurologically intact participants

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	p-value for linear fits
Transition Vertical Velocity Difference	Linear	0.228 ± 0.038	-0.002 ± 0.010	0.564	1.44e-21*
CoM Acceleration at initial contact	Power $ax^b + c$	a:0.499 ± 0.225 b:1.983 ± 0.777	c: -0.026 ± 0.016	0.556	

Centre of Mass (CoM), *significant result for linear fit

The measures describing CoP intersection variability in the butterfly diagram did not show strong relationships to walking speed. The variability in the ML direction showed the strongest relationship, and this is potentially related to the similar relationships seen in step width variability and lateral impulse.

Table 3-4: Regression results for butterfly diagram characteristics with respect to walking speed in neurologically intact participants

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	P
ML CoP Intersect SD	Linear	-0.027 ± 0.016	0.028 ± 0.004	0.249	0.002*
AP CoP Intersect SD	Linear	-0.011 ± 0.033	0.036 ± 0.009	0.007	0.528
CoP Intersection Density	Linear	268 ± 184	118 ± 48	0.137	0.005

Medial-lateral (ML), anterior-posterior (AP), Centre of Pressure (CoP), standard deviation (SD), *significant result for linear fit

Within individual trials, the standard deviation was calculated to provide an estimate of variability across the 15 stride samples. The variability of CoM acceleration at initial contact and velocity change over double support showed increases with walking speed. The

variability of temporal metrics such as double support time and duty factor of each leg decreased with walking speed.

3.2.2. Parameters without speed dependence

For parameters which showed no discernable relationship to walking speed, the means and standard deviation are shown in Table 3-5 below. These are generally measures of symmetry and AP direction variability.

Table 3-5: Neurologically intact participant mean values and standard deviations for non speed dependent metrics

Parameter	Mean (SD)
Difference in Acceleration at Initial Contacts	0.0186 (±0.0189)
Lateral CoM Velocity Range	0.128 (± 0.023)
Lateral CoM Velocity Asymmetry*	0.509 (±0.007)
Lateral Impulse Variability	0.004 (± 0.001)
Step Length Variability	0.021 (± 0.010)
Step Width	0.238 (± 0.036)
Difference in Max Forward CoM Velocity	0.009 (± 0.006)
Step Length Asymmetry*	0.513 (± 0.010)
Propulsive Impulse Ratio*	0.452 (± 0.057)

Centre of Mass (CoM), *Asymmetry values listed have perfect symmetry at a value of 0.5

3.3. Discussion

Many of the known relationships with walking speed are maintained at slower speeds, including step length, double support time, and cadence. This confirms the results of Wu et al. (2019) which used a smaller sample size but used both kinetic and kinematic measures of gait. The mean step width was higher in this data, most likely due to all participants being naïve to split-belt treadmills to meet the inclusion criteria of the related pilot. Increased step width on a split-belt is common for new users, as they are often told to avoid crossing the

midline of the treadmill (Finley, Long, Bastian, & Torres-Oviedo, 2015). The consistency of the regression results with previous literature at different speeds gives credence to the data obtained from an instrumented treadmill alone with no kinematic assistance.

The increase in lateral impulse with slower speeds is not reported in the previous studies but could correspond to the increase in ankle torque seen at reduced speeds in slow walking (Best, 2020). This also corresponds to the need for active stabilization in the lateral direction and the increase in lateral instability at slower speeds (Bauby & Kuo, 2000).

Symmetry is often an inherent assumption made for neurologically intact gait data (Adamcyzk, 2008); our results confirm this across most measures. The propulsive impulse ratio shows minor asymmetry, with a mean of 0.452 for the neurologically intact individuals. This could possibly be attributed to the effect of a dominant leg, as suggested to explain asymmetry in previous studies (Herzog, Nigg, Read, & Olsson, 1989). This range for asymmetry in neurologically intact individuals helps provide context for the stroke data presented in chapter 4.

The increased variability present at slower walking speeds in neurologically intact individuals could indicate that they are less able to exploit the passive dynamic aspects of walking. Passive dynamic aspects of walking have been proven to allow human like motion in non-powered bipedal robots (Collins et al., 2001). Where features such as arm swing and knee joints allowed for the robot to walk downhill without any additional actuation. The stability of passive dynamics is often likened to riding a bicycle, where there is a natural stability at higher speeds compared to the wobbling motion which causes difficulties at lower speeds. There is potential that human movement is less passively stable at these very slow speeds.

3.3.1. Neurologically intact gait hodographs

We can visualize the CoM metric results by using the corresponding hodographs. Figure 3-3 shows a single stride hodograph from a neurologically intact participant at self-selected walking speed. The characteristic shape is common for neurologically intact participants

around average comfortable walking speeds. A figure-8 in the frontal plane, saddle shape in the coronal, and ‘D’ shape in the sagittal plane. The trajectory of the hodographs is consistent between strides at higher speed, with more variability introduced at slower speeds.

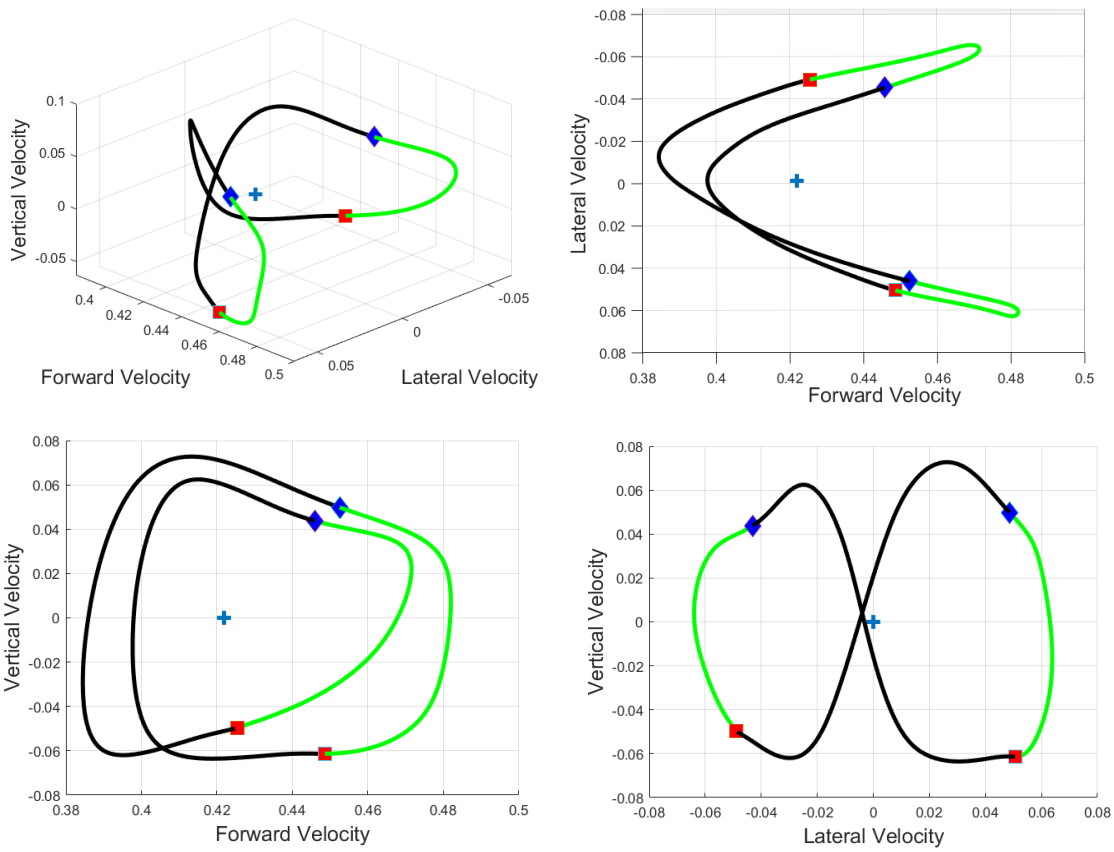

Figure 3-3: Single stride hodograph for a neurologically intact participant
Clockwise from top: isometric view, top view with walking direction to the right, frontal plane viewed from behind, sagittal plane view with walking direction to the right. Red squares indicate initial contact, blue diamonds indicate foot-off, black lines are single support, green lines are double support. The plus indicates the mean velocity, equivalent to treadmill speed in forward direction.

In neurologically intact individuals the overall shape of the hodograph follows one of two trends as velocity decreases. Figure 3-4 shows the variation where the vertical velocity difference between initial contact and foot-off decreases, and the hodograph flattens while remaining laterally symmetrical. The vertical movement of the CoM is limited as speed decreases.

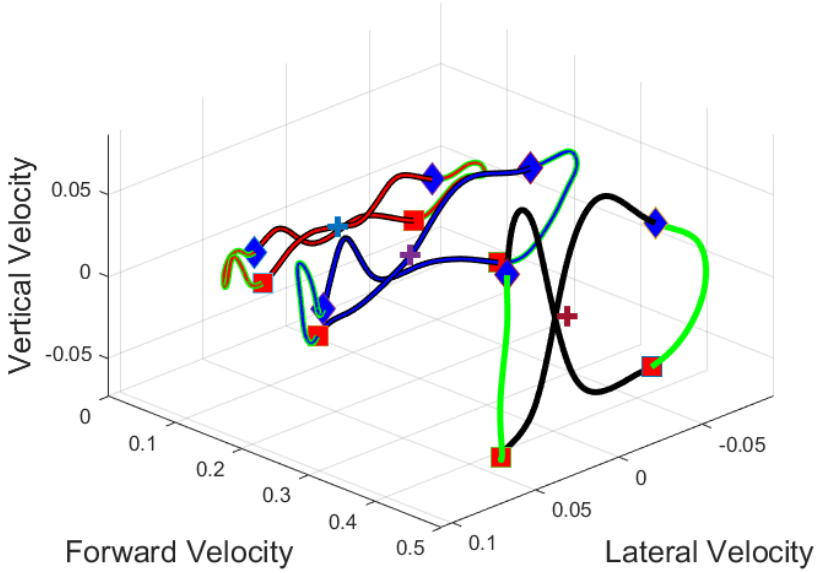

Figure 3-4: 3D hodographs flattening with decreasing walking speed

The other pattern seen in Figure 3-5 is an inversion of the velocity change between initial contact and foot-off on one side. This pattern is characterized by a round trajectory in the frontal plane and correlates with reduced vertical ground reaction force during the step-to-step transition on one side. This can be seen as a gait that is stilted to one side and lingers in double support time. The impact and cause of these different patterns in neurologically intact participants was not investigated. The unfamiliarity of walking at such slow speeds could have prompted the difference in strategies.

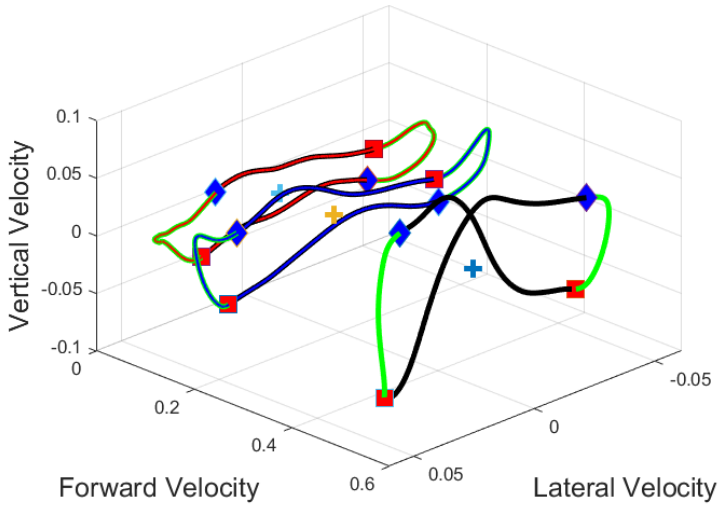

Figure 3-5: 3D hodograph unilaterally inverting with decreasing walking speed

3.3.2. Butterfly diagrams for neurologically intact gait

The butterfly diagrams seen in Figure 3-6 demonstrate how the forward range of the CoP decreases with walking speed as step length decreases. The lateral range of CoP is not affected by speed, reinforcing our results regarding step width. The variability in the intersection points can be interpreted as either variation in foot placement if the proportions of the butterfly diagram in the direction of travel are also varied, or as variability in weight shift if the rest of the plot is consistent.

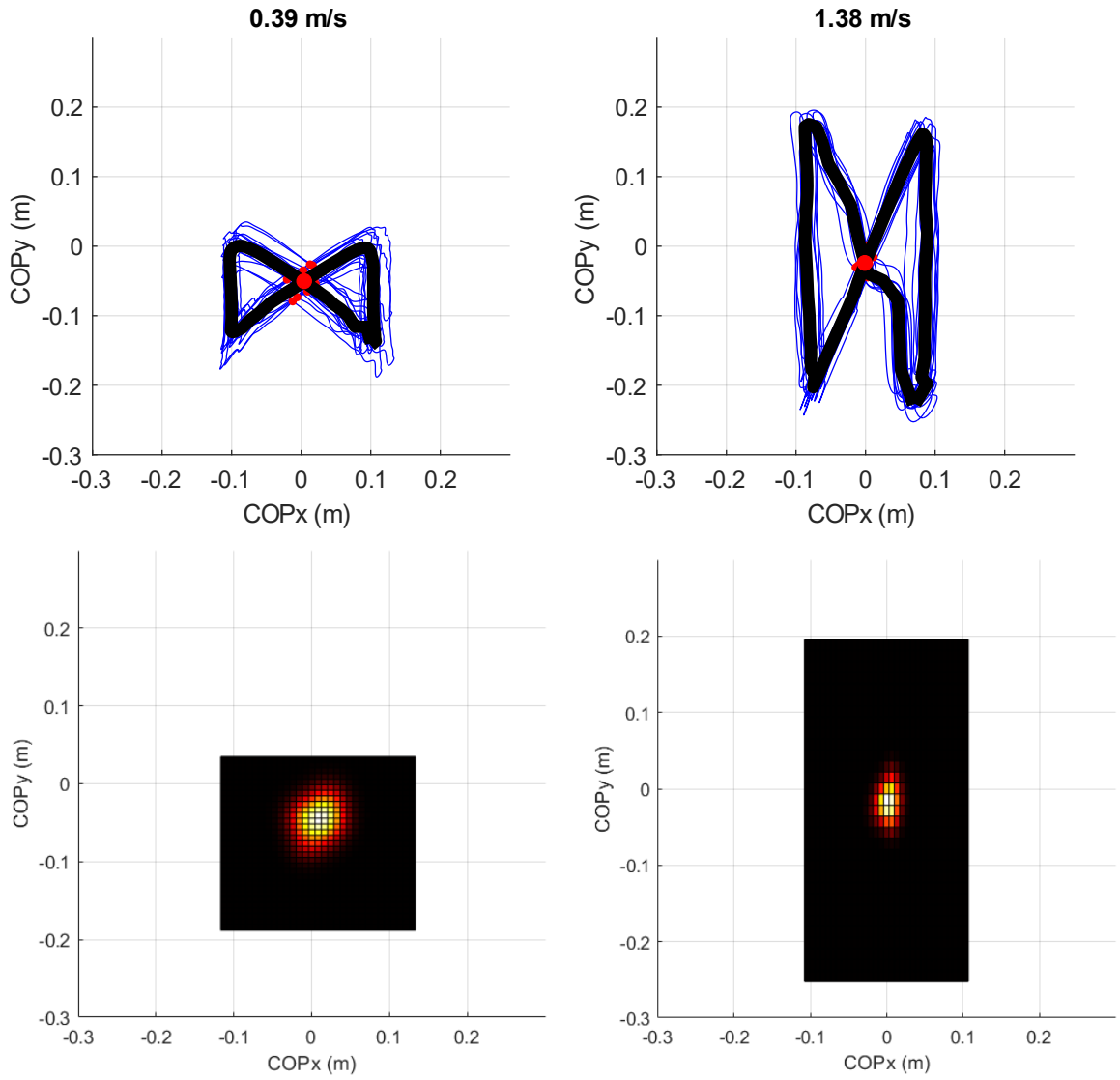

Figure 3-6: Butterfly diagrams for neurologically intact gait at slow speed vs self-selected speed

Top: Butterfly diagrams for neurologically intact participant. Blue lines indicate the CoP trace for individual strides, the black line is the mean. Red dots indicate the location of intersections. Bottom: Corresponding kernel density estimation plots

Chapter 4. Kinetic gait analysis of subacute stroke patients during inpatient rehabilitation

4.1. Methods

4.1.1. Participants

Participants were recruited as part of a larger study on the effect of split-belt treadmill training in subacute stroke rehabilitation. 15 stroke patients (8 female, 7 male) were recruited from the Neuro Rehab Unit of the Foothills Hospital in Calgary, Alberta. Potential participants were identified by relevant clinicians working on the unit. The study team approached potential participants for recruitment and went through the informed consent process with them. The inclusion criteria for the study required that participants were age 18 or older and had recently suffered their first neurological injury. Participants were excluded if they were unable to participate in treadmill walking due to ongoing comorbidities determined by the clinical rehabilitation teams. Participants exhibited a variety of deficits including hemineglect, aphasia, and spasticity.

Data collection coincided with ongoing rehabilitation sessions. The patient's physiotherapist was present for all data collection to not interfere with the ongoing rehab. At baseline all participants were within 6 months poststroke, with one participant crossing the 6-month mark during data collection. The mean (±SD) age of participants was 55 (±14) years old, mean (±SD) body mass was 79 (±25) kg, with a mean (±SD) leg length of 82.0 ± 7.0 cm. The mean (±SD) time post stroke at first assessment was 78 (±57) days. The mean (±SD) comfortable walking speed of participants prior to the first session measured by the 10MWT was 0.53 (±0.38) m/s (0.19 ± 0.13 non dimensional). The basic demographics, type/side of stroke, and assessment scores are located in Appendix A.

4.1.2. Protocol

Baseline assessments were conducted prior to the first session of treadmill training. The 10MWT was used to establish initial walking speed, 6MWT assessed endurance, and

the Fugl-Meyer Lower Extremity assessment (FM-LE) as a clinical baseline. The FM-LE is reported as a score out of 34 for each leg, measuring aspects of joint motion, pain, sensation, coordination, reflexes, and voluntary movements in a variety of positions (Fugl-Meyer, Jaasko, Leyman, Olsson, & Steglind, 1975). Participants were able to use gait aids (canes, walkers, or ankle foot orthosis (AFO)) during both the 10MWT and 6MWT.

Heart rate and blood pressure were measured before treadmill sessions and between walking bouts to ensure patient safety. During treadmill sessions, all patients wore a harness attached to an overhead rail to prevent falls in case of loss of balance. Based on physiotherapist recommendation, some patients were provided with low amounts of body-weight support if required. Most participants required the use of a handrail located at the front of the treadmill with either one or both hands. The impact of handrail and body-weight support is quantified as the average impulse that would be required to complete the limit cycle of the hodographs. After this value is calculated, the effect of external assistance is offset in the velocity calculations.

Each session of treadmill training began with at least 2 minutes of tied belt walking at comfortable speed. This speed was determined by the participant and their physiotherapist. This resulted in multiple speeds being recorded for some sessions as participants were able to increase their comfortable speed over the course of rehabilitation. Some participants then underwent split-belt treadmill training as part of a related pilot study, while others continued to use tied-belt for the duration of the rehabilitation sessions.

The number of treadmill sessions varied between participants, ranging from 1-13 sessions, with a mean and median of 6 sessions across all participants. Those who completed a larger number of sessions were scheduled for up to 3 sessions a week, and after the 3rd week of sessions a second set of assessments was performed.

The protocol was conducted in a way that interfered minimally with the rehabilitation that the participants received. All sessions were attended by one or more of their inpatient rehabilitation physiotherapists. At some points during treadmill training, physiotherapists

coached or facilitated the gait of some participants. Participants were able to use the arm slings and AFOs given to them at the recommendation of their physiotherapist.

4.2. Results

4.2.1. Stroke gait correlations with non dimensional walking speed

Fits for the poststroke walking data, including fits to paretic and nonparetic legs individually are shown in the following tables. The Bonferroni correction for significance was used to account for multiple tests ($\alpha = 0.002$). The general quality of the fits is lower compared to the neurologically intact gait data, highlighting the variability of gait poststroke.

Table 4-1: Spatiotemporal fits with respect to walking speed in stroke

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	Linear fit P-value
Step Length Paretic	Power ax^b	a: 1.429 ± 0.235 b: 0.597 ± 0.088		0.597	
Step Length Nonparetic	Power ax^b	a: 1.390 ± 0.141 b: 0.673 ± 0.056		0.800	
Step Width	Linear	-0.479 ± 0.128	0.385 ± 0.020	0.264	9.3e-12*
Step Width SD	Linear	-0.045 ± 0.032	0.033 ± 0.005	0.047	0.007
Double Support Time (% stride)	Power ax^b	a: 0.202 ± 0.017 b: -0.384± 0.036		0.737	
DST Paretic Leading (% stride)	Power ax^b	a: 0.078 ± 0.012 b: -0.472± 0.062		0.600	
DST Nonparetic Leading (% stride)	Power ax^b	a: 0.129 ± 0.021 b: -0.302± 0.065		0.345	
Cadence (Hz)	Power ax^b	a: 2.292 ± 0.307 b: 0.302 ± 0.065		0.370	

Standard deviation (SD), double support Time (DST), Hertz (Hz), *significant result for linear fit

Table 4-1 shows that there are slight variations to the fits for paretic vs nonparetic sides in step length, and double support time. Nonparetic step length and paretic double support time show the stronger relationship with increasing speeds compared to their counterparts. Step width shows a decrease as velocity increases, while step width variability does not show a strong relationship to walking speed.

Table 4-2: Force and impulse fits with respect to walking speed in stroke

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	P
Paretic Propulsive Impulse (BW*s)	Linear	0.061 ± 0.023	0.007 ± 0.003	0.160	2.8e-07*
Nonparetic Propulsive Impulse (BW*s)	Linear	-0.165 ± 0.054	0.071 ± 0.008	0.198	7.52e-09*
Propulsive Impulse Ratio	Linear	1.265 ± 0.260	0.088 ± 0.040	0.379	2.0e-17*
Paretic Lateral Impulse (BW*s)	Linear	-0.059 ± 0.041	0.064 ± 0.007	0.051	0.005
Nonparetic Lateral Impulse (BW*s)	Linear	-0.242 ± 0.058	0.096 ± 0.009	0.303	1.3e-13*
Paretic Peak Propulsive Force (BW)	Linear	0.266 ± 0.043	0.020 ± 0.006	0.502	8.4e-25*
Nonparetic Peak Propulsive Force (BW)	Linear	0.299 ± 0.044	0.051 ± 0.006	0.539	2.3e-27*
Mean Vertical Force in Paretic Leading DS (BW)	Linear	0.289 ± 0.089	0.974 ± 0.014	0.212	1.9e-9*
Mean Vertical Force in Nonparetic Leading DS (BW)	Linear	0.276 ± 0.077	1.021 ± 0.012	0.251	3.7e-11*

Double support (DS), bodyweight (BW), *significant result for linear fit

The nonparetic propulsive impulse shows a decrease with increasing walking speed, the reverse of what is seen in the paretic leg and neurologically intact control. The ratio between paretic and nonparetic impulses shows a stronger relationship to walking speed than its individual components. Lateral impulse generated by the nonparetic leg shows a stronger relationship with walking speed than the paretic leg, with a greater negative slope than both paretic and neurologically intact.

The CoM trajectory characteristics do not show strong trends related to walking speed in the stroke group as a whole. Of the metrics examined, the difference in velocity across the transition with the paretic leg leading shows the strongest fit to walking speed. Only paretic leading change in velocity and CoM acceleration at nonparetic initial contact showed significant relationship to walking speed.

Table 4-3: CoM trajectory characteristic fits to walking speed in stroke

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	P
Paretic Leading Vertical Velocity Difference	Linear	0.302 ± 0.111	-0.040 ± 0.017	0.159	3.0e-7*
Nonparetic Leading Vertical Velocity Difference	Linear	-0.150 ± 0.116	0.109 ± 0.018	0.040	0.012
CoM Acceleration at Paretic IC	Linear	-0.045 ± 0.100	-0.072 ± 0.016	0.005	0.371
CoM Acceleration at Nonparetic IC	Linear	-0.198 ± 0.116	-0.030 ± 0.018	0.07	0.0008*

Centre of mass (CoM), initial contact (IC), *significant result for linear fit

The ML CoP intersection variability shows the strongest correlation with walking speed of the CoP metrics examined. The variance explained by walking speed for ML variability is a stronger fit in the stroke data (R^2=0.457) compared to the neurologically intact data (R^2=0.249). The AP variability shows no relation with speed, the same as was seen in the neurologically intact data.

Table 4-4: CoP intersection fits with respect to walking speed in stroke

Metric	Fit Type	Coefficient (±95%CI)	Offset	R^2	P
ML CoP Intersect variability	Linear	-0.139 ± 0.030	0.045 ± 0.005	0.357	4.91e-17*
AP CoP Intersect variability	Linear	0.007 ± 0.027	0.024 ± 0.004	0.001	0.605
Maximum KDE (m^{-2})	Linear	720 ± 253	96.6 ± 38.3	0.186	0.0001*

Medial-lateral (ML), anterior-posterior (AP), kernel density estimation (KDE), *significant result for linear fit

4.2.2. Comparison of speed relationships between neurologically intact and poststroke

The relationship between the two groups' gait metrics and walking speed was compared using multiple linear regression. This analysis was only completed on variables which had linear relationships with walking speed in at least one of the groups. The assumptions of normality and independence of observations are not valid for these comparisons as each individual had their gait metrics calculated for multiple walking speeds. For unilateral metrics, the neurologically intact gait metrics were compared to the nonparetic and paretic side data individually. The null hypothesis for these comparisons being that the relationship to walking speed has the same slope in both groups. The same level of significance ($\alpha = 0.002$) is used following a Bonferroni correction. The significance testing for the comparison of different slopes is found in Table 4-5.

Table 4-5: Multiple linear regression results for group comparison

Variable	Side	p-value for difference in slopes
Peak Propulsive Force (BW)	Paretic	9.19e-13*
	Nonparetic	9.73e-8*
Propulsive Impulse (BW*s)	Paretic	0.011
	Nonparetic	4.21e-9*
Propulsive Impulse Ratio		2.77e-13*
Lateral Impulse (BW)	Paretic	0.707
	Nonparetic	4.60e-8*
CoM Acceleration at Initial Contact (g)	Paretic	8.80e-8*
	Nonparetic	1.9e-10*
ML CoP Intersect variability		1.48e-10*
Maximum KDE (m^{-2})		0.062
Transition CoM vertical velocity difference	Paretic	0.25
	Nonparetic	5.37e-8*

Bodyweight (BW), center of mass (CoM), acceleration of gravity (g), medial-lateral direction (ML), center of pressure (CoP), kernel density estimation (KDE), *significant difference between slopes at 0.002 confidence level

4.2.3. Measures of gait recovery

Four participants were able to attend 8-13 sessions each over the course of 1-2 months. The demographics, clinical scores, and 3-week follow-up scores are shown in Table 4-6. Participant GAQPQ crossed 6 months poststroke during rehabilitation, their data is presented to provide reference for recovery outside of the early rehabilitation stages. All participants showed improvements in endurance measured by the distance traveled in the 6MWT. All early recovery participants showed improvements in the Fugl-Meyer LE scores, the later stage participant remained the same. All participants saw increases to their 10MWT walking speed, with the exception of one participant who took the test using a 4-wheel walker initially and saw a decrease in walking speed after transitioning to a quad cane.

Table 4-6: Demographics and clinical measures for long-term participants

ID	Age	Sex	Days Since Stroke at First Assessment	Number of Session Before 3-Week Follow-up	Initial and 3-Week FM-LE (Left-Right)	10MWT m/s (ND)	6MWT (m)
CUEGK	83	M	16	7	34-19 34-26	0.46 (0.154) 0.64 (0.214)	137 189
GGOOV	51	F	54	9	34-22 34-27	0.476 (0.165) 0.90 (0.313)	136 280
BUQBS	51	F	73	8	34-19 34-25	0.23 (0.084) 0.17 (0.062) *	84 98
GAQPQ	75	F	174	9	15-34 15-34	0.18 (0.065) 0.23 (0.083)	55 66

Fugl Meyer Lower Extremity Assessment (FM-LE) has a maximum score of 34 for each side, 10-meter walk test (10MWT), nondimensional walking speed (ND) in brackets, 6-minute walk test (6MWT), *Participant completed the 10MWT with a 4-wheel walker in first session, moved to quad cane for follow-up

Table 4-7: Deficits and equipment used during treadmill sessions for long-term participants

ID	Aphasia, visual, or attention deficits	Equipment used on treadmill
CUEGK		Bodyweight support, handrail
GGOOV	Expressive aphasia, right hemineglect	Bodyweight support, handrail, AFO*
BUQBS		Bodyweight support, one handed handrail, arm sling
GAQPQ		Bodyweight support, one handed handrail, arm sling, AFO*

*Ankle foot orthosis (AFO) brand and type not recorded

Figure 4-1 shows the progression of walking speed over the course of treadmill training. The initial treadmill walking speed in the first sessions was lower than the 10MWT results in initial evaluations. This could be due to the unfamiliarity with the system. All participants showed improvements in walking speed on the treadmill. The participants with lesser initial impairments showed greater improvements in walking speed. The near-chronic participant showed the least improvement.

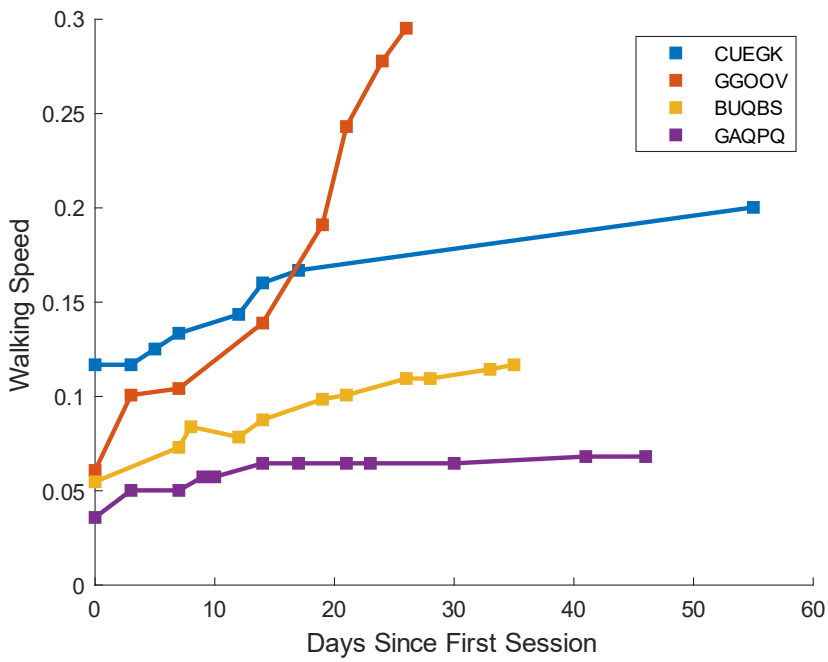

Figure 4-1: Nondimensional walking speed for participants during treadmill training

There were degrees of variability between consecutive sessions for each metric depending on the metric and participant. For the remainder of the plots in this section a moving average trendline is shown to ease interpretation of the overall trends in the gait metrics. The average value of each metric in a session is shown.

Figure 4-2 depicts changes in cadence over the course of rehabilitation. The older participants (age 75 & 83) show a tendency towards a lower cadence compared to the younger participants (both age 51). Double support time shown in Figure 4-3 decreases as expected,

with the participants who achieved the faster walking speeds showing a lower value consistent with the relationship seen in neurologically intact individuals at those speeds.

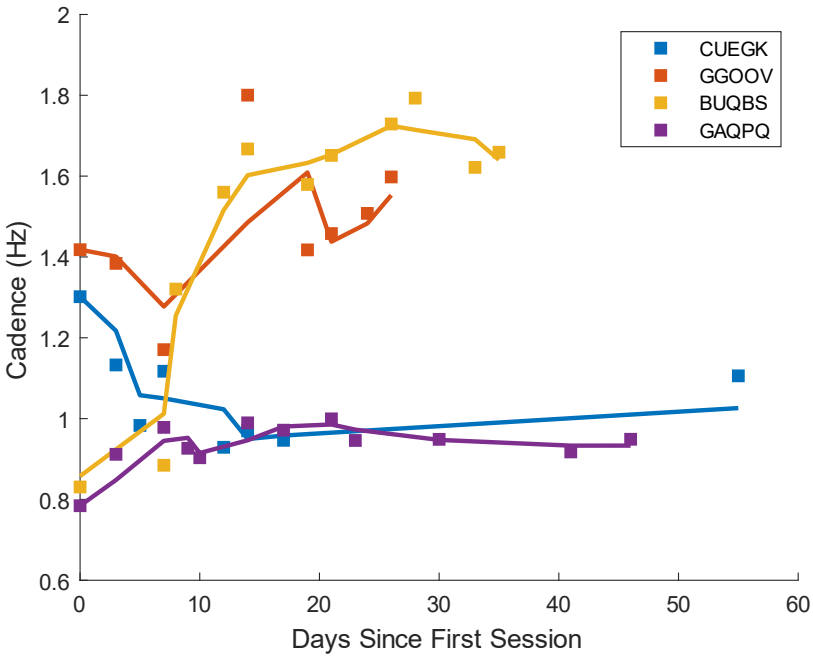

Figure 4-2: Cadence vs rehabilitation time

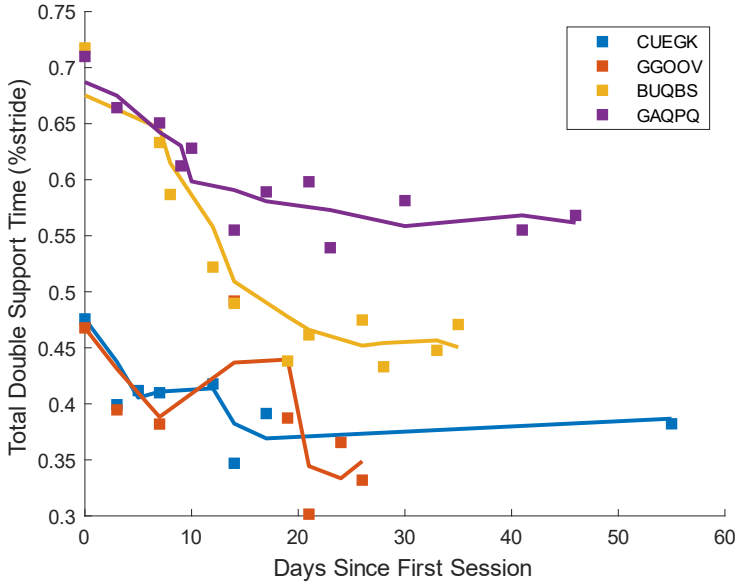

Figure 4-3: Double support time vs rehabilitation time

The duty factor for paretic and non-paretic legs is shown in Figure 4-4. All participants show asymmetry in the amount of time spent in stance in favour of the nonparetic leg. This corresponds to reduced swing time for the nonparetic leg.

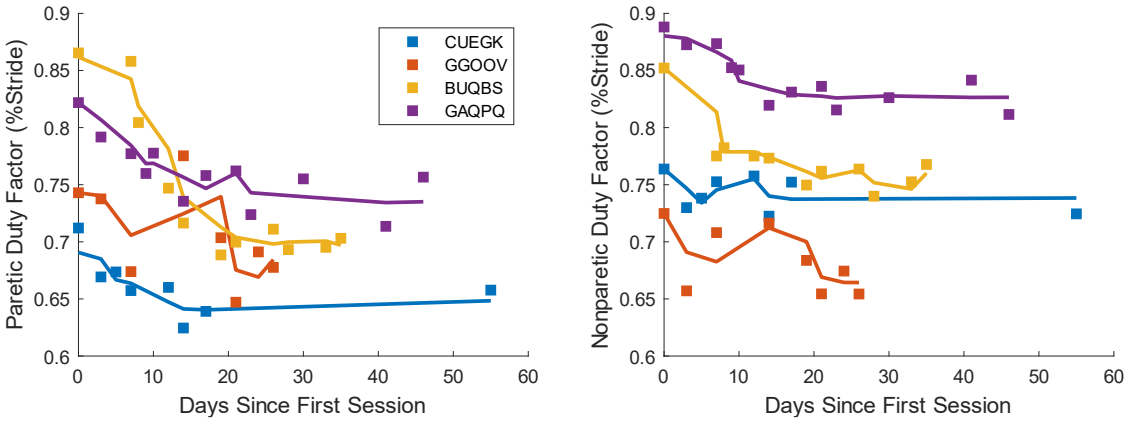

Figure 4-4: Comparison of duty factor (stance time as %stride)

Figure 4-5 depicts the change in step length asymmetry. The two older group together again with a mildly asymmetric step length (~0.54), while the younger pair show more variance but are closer to symmetric gait (0.5).

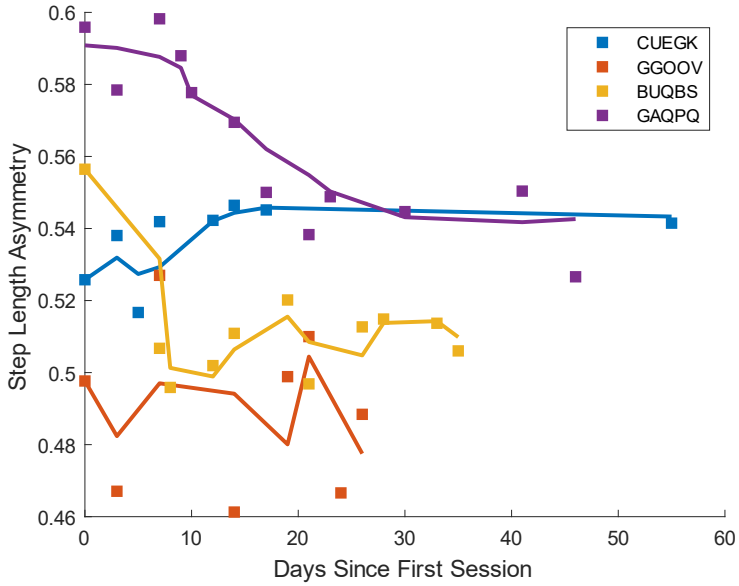

Figure 4-5: Step length asymmetry vs rehabilitation time

Figure 4-6 shows that only one participant saw an increase in propulsive impulse ratio while the remaining 3 remained similar over time. The near-chronic participant showed the largest asymmetry in propulsive impulse. They were unable to walk without using the handrail and only used one hand on the rail. The participant who achieved more symmetric propulsion saw a bilateral change in propulsive impulse, reducing their non paretic output and increasing the output of the paretic side. Figure 4-7 shows that all participants reduced the variability of their propulsive impulse, with a more consistent end point seen in the paretic leg.

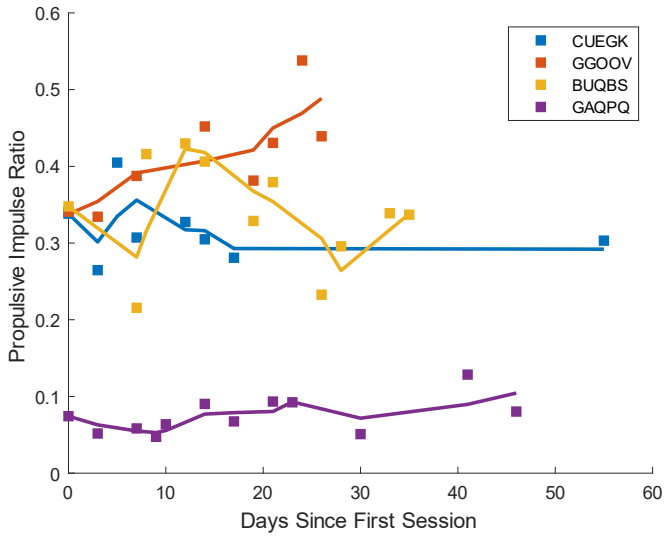

Figure 4-6: Propulsive impulse ratio vs rehabilitation time

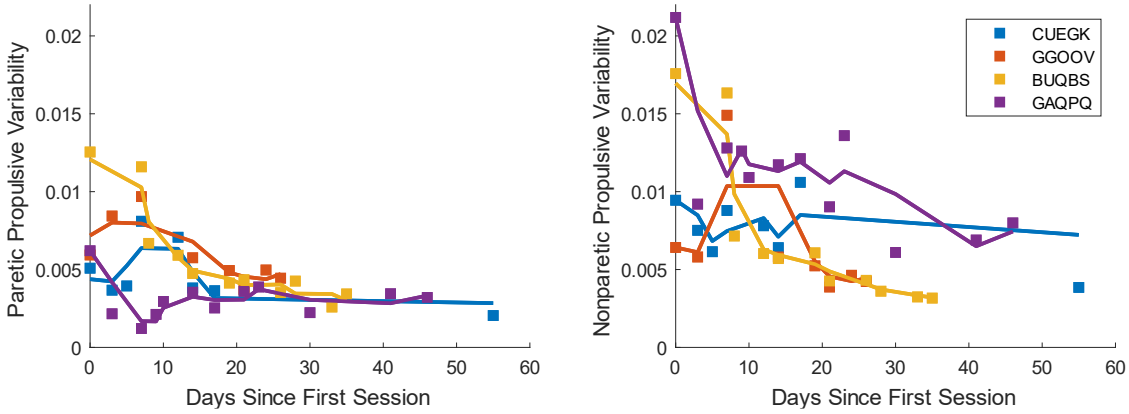

Figure 4-7: Variability (SD) of propulsive impulse vs rehabilitation time

Example hodographs for two timepoints are shown in Figure 4-8. The lateral tilt in the frontal plane is a common feature seen in stroke participants who use hip-hiking, a strategy that involves tilting the pelvis to increase foot clearance on one side.

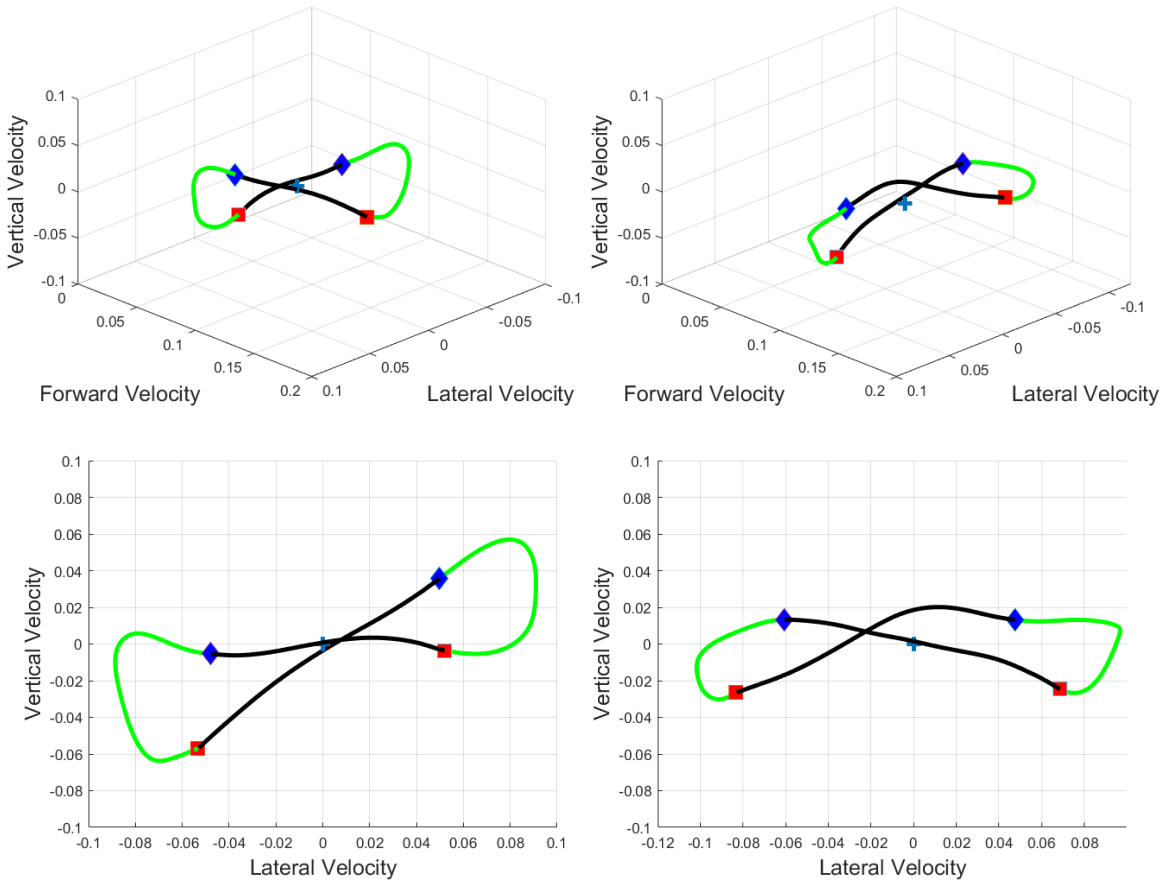

Figure 4-8: Hodographs for stroke participant at 2 and 5 week timepoints
Top: 3D views of hodographs for participant BUQBS at 94 days (Left) and 113 days (Right) Bottom: Corresponding frontal plane hodographs for the same stride. Of note is the shift in orientation of the lateral axis to a more symmetric pattern of vertical velocity.

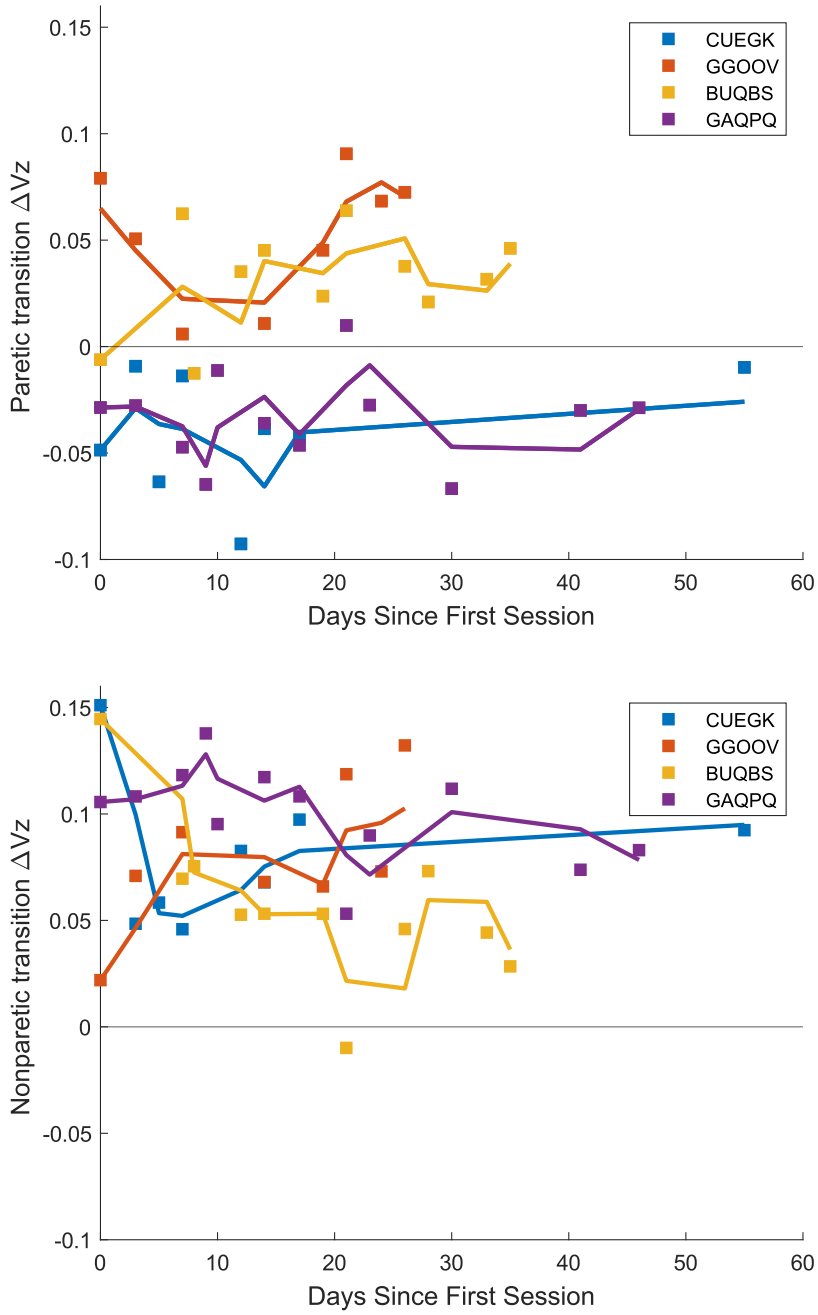

Figure 4-9: Change in vertical velocity during step to step transition
Equivalent to the vertical distance between initial contact and foot-off in the hodographs. The negative difference on the paretic side in the older participants indicates that they would have inverted hodographs similar to those in Figure 3-5.

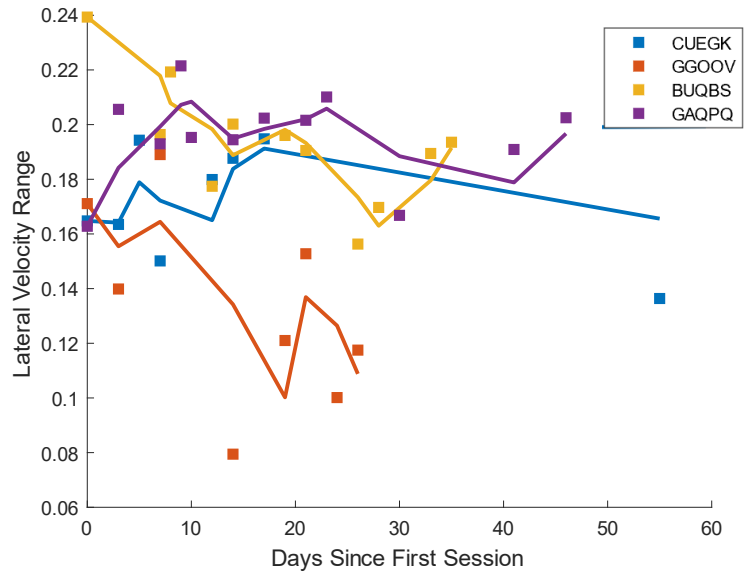

Figure 4-10: Lateral CoM velocity range

Equivalent to the width of the hodograph in the frontal plane. Shows a consistent decrease across all participants with time.

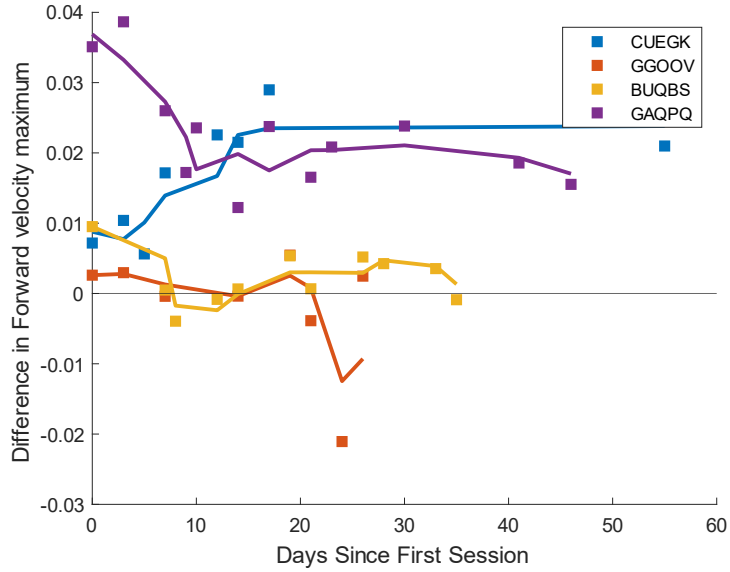

Figure 4-11: Forward CoM velocity symmetry

Equivalent to the absolute difference in forward velocity between the paretic leading step and the nonparetic leading step. The positive value indicates the nonparetic push off while the paretic leg is leading achieves a higher CoM velocity than the opposing step.

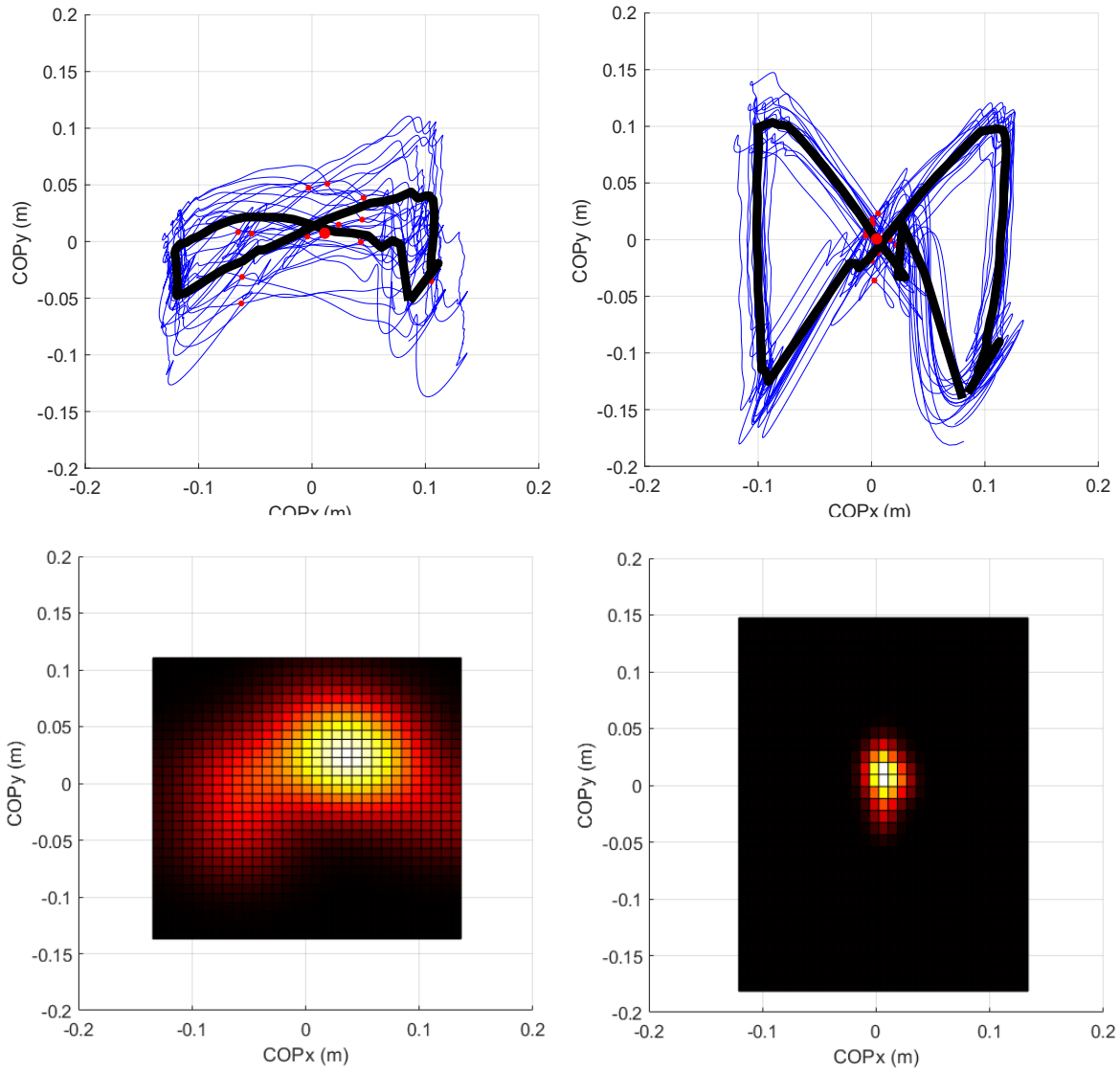

Figure 4-12: Butterfly diagrams for a single stroke participant at initial and 3 week timepoints

Top: Butterfly diagrams for participant GGOOV at 57 days (Left) and 78 days (Right) poststroke. Blue lines are consecutive stride CoP traces. Red dots are intersections. Black line is mean CoP trace. Bottom: Corresponding KDE plots illustrating the decrease in intersect variability. White indicates the location of the highest intersection density.

The variability in CoP intersections is shown in Figure 4-13 for the AP direction and Figure 4-14 for the ML direction. The variability in both direction shows a trend of decreasing over time across all participants. The AP variability appears to plateau earlier (~10 days) compared to the ML variability which continues to decrease after 20 days of training.

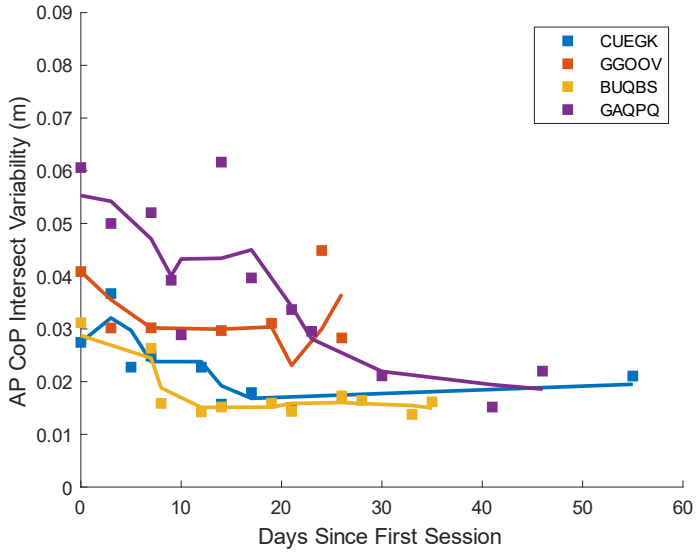

Figure 4-13: AP variability of CoP intersection in butterfly diagram

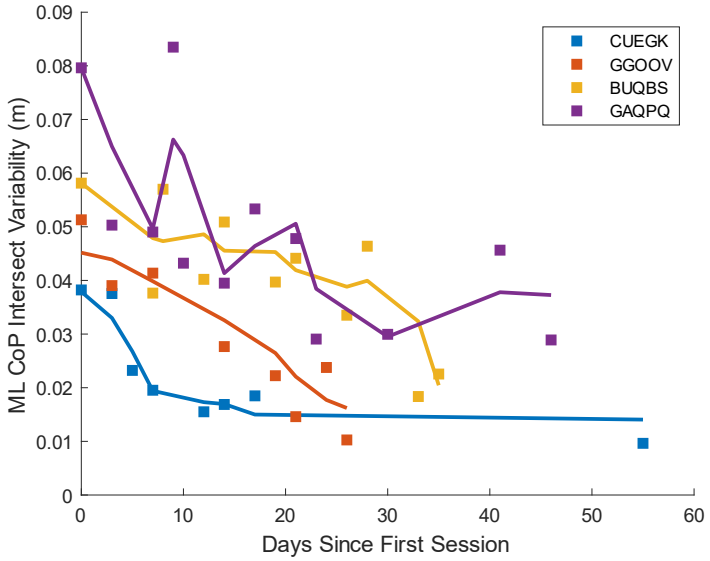

Figure 4-14: ML variability of CoP intersection in butterfly diagram

The KDE analysis shows an increase in the intersection density across the rehabilitation period in all participants. There are several points which could be considered outliers compared to the trends, and the two younger participants showed large variability between consecutive sessions.

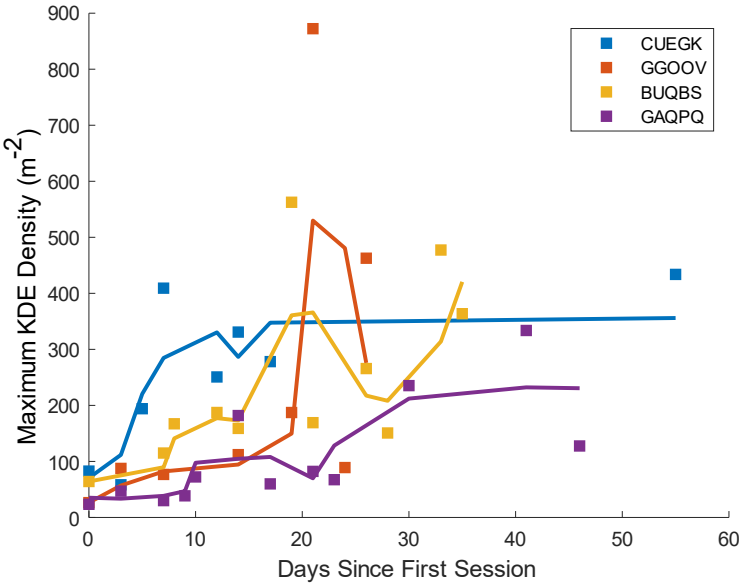

Figure 4-15: Kernel density estimation (KDE) of CoP intersections in the butterfly diagram

Maximum kernel density of the CoP intersections in the butterfly diagram measured by the methods described in (Lee & Liang, 2020).

4.3. Discussion

4.3.1. Changes in speed relationships compared to neurologically intact controls

The stroke data shows changes in the relationship to walking speed across many variables. Step width saw a decrease with respect to walking speed poststroke but not in the neurologically intact data. This could be due to slower poststroke participants having higher instability and adopting a larger base of support. As the poststroke participants increase in

speed the wide stance becomes less efficient and therefore a decrease in step width may be warranted.

Step width variability showed a decrease with walking speed poststroke compared to an increase in controls. Previous studies correlated step width variability with increased active balance control (Bauby & Kuo, 2000). This could potentially indicate that lower speeds increase instability and require more active modulation of step width in order to compensate.

The total double support generally followed the relationship seen in neurologically intact gait. The individual step double support time showed a mild asymmetry in the walking speed relationships between left and right leading steps. The paretic leading step showed a stronger relationship to walking speed, indicating that lingering in double support before moving to single stance on the paretic leg may act as a rate limiting factor for walking speed.

Propulsive impulse ratio showed a strong difference between stroke and controls as expected. This presents in the individual leg relationships as opposite signs on the slope of walking speed fits. This indicates that slower poststroke participants use more nonparetic compensation while faster poststroke participants trend towards symmetry.

The nonparetic leg shows larger impulses than the paretic or control results at similar speeds. This indicates that the nonparetic leg may take on a larger role in stability via lateral foot placement or ankle torques, similar to the idea proposed for slow walking in Best (2020).

Propulsive peak forces showed weaker fits to walking speed poststroke than in neurologically intact controls but were still relatively strong in their relationship. Both paretic and non paretic legs showed a reduced slope in the fit compared to the control data. This may indicate that the ability to generate propulsive forces is either impaired, or potentially the asymmetry in the ability to generate such forces would cause their use to be destabilizing overall.

Mean vertical force showed an asymmetry in the intercept between paretic and nonparetic leading double support. The paretic side intercept goes below one body-weight, while the nonparetic side remains greater than one body-weight. This pattern when seen in controls results in the partially inverted version of hodographs seen in Figure 3-5, where vertical velocity decreases during double support on one side vs increasing on the opposing side. The difference in vertical velocity during double support shows a negative intercept in the fit for the paretic side and a positive intercept on the nonparetic. This indicates that the inversion of the hodograph at these slow speeds would be on the paretic leading step. This relationship, which is seen in both poststroke participants and neurologically intact controls, may indicate that at slow speeds there is some benefit to an asymmetric change in vertical velocity. This could indicate that there is a priority to have the nonparetic or dominant leg in a more optimal initial condition at initial contact. Whether this strategy is for a stability or energetic benefit is unknown.

The acceleration of the CoM at initial contact showed a good relationship to walking speed in neurologically intact controls. This was not present in the poststroke participants, who had negative values for acceleration at initial contact for both legs across all trials. This indicates that there is reduced pre-emptive push off, leading to more energetic losses in the collisions. At the slowest speeds in controls there were negative accelerations at initial contact for a few of the participants. This may indicate that at these speeds, the slower collisions cause limited losses to justify the energetic trade off for reducing collisions with pre-emptive push-off. The lack of positive acceleration at initial contact across all poststroke trials, even those at increased speeds, may indicate that there is an impairment to the generation or timing of push-off.

The ML variability of butterfly plots is increased at slower speeds poststroke. This can be indicative of inattention or drifting horizontally while walking or by variability in weight shift between steps. The stronger relationship with walking speed seen in stroke participants indicates that the ML stability during walking may be a factor in the limited speed of individuals poststroke. KDE showed a weaker relationship in both control and

poststroke groups. This could be due to it combining the variability of both AP and ML directions into a single metric which resulted in a fit between the two in quality.

4.3.2. Individual recovery

Examining the longitudinal recovery of the subset of poststroke patients gave a number of insights that would not have been apparent from the group data alone. Different gait parameters showed commonalities in recovery between participants with a shared aspect. These groups could be divided among aspects including the age differences, higher final speeds, borderline chronic vs subacute, and other potential sources of covariation in recovery.

The relationship between cadence, step length, and velocity explored in previous studies (Bertram, 2005) appear to be maintained here albeit likely with a different cost surface. The asymmetries in various spatiotemporal measures are likely the result of other parameters rather than the cause.

The results for propulsive impulse ratio were surprising when compared to the group effect. It was expected that a strong relationship to walking speed would be observed and the ratio would improve towards symmetry, but only one of the long-term participants showed any strong improvement. There is potential that the use of handrails impacted this result, especially in the near chronic case. The decrease in impulse variability could be interpreted as the gait pattern becoming more consistent with recovery and training.

Parameters derived from CoM velocity and acceleration showed the impact of the combinations of the other gait parameters. The vertical force differences in double stance relate to the change in vertical velocity across the step-to-step transition. The lateral velocity range reflects increases in lateral stability. The forward velocity differences reflect the asymmetry in propulsion.

The CoP variability metrics seem to be primarily related to aspects of stability. With an increase in stability with recovery explaining the results across each measure. Overall, the ML variability seems to be the key measure among CoP values. It shows strong relationships

to walking speed in both control and stroke groups as well as improvements in the recovery subset. These improvements may highlight the active stability requirements of slow walking when the normal passive dynamics of gait are less present and beneficial.

4.3.3. Gait visualizations as a tool in rehabilitation

The characteristic shape of the hodograph in neurologically intact individuals is immediately distinguishable from the distorted shape found early poststroke. The summation of all gait deviations presents their effects on the trajectory of the CoM. Certain aspects of the shape are unique to the individual impairments seen in stroke participants. For example, a participant with a stomping gait showed very sharp corners in the hodograph at initial contact. This indicates a large amount of energy is lost in the collisions which immediately redirect the CoM. These individualities and eccentricities in the shapes of the hodographs match well to the observations given by the physiotherapists during sessions. Matching observations to the hodographs in future rehabilitation sessions could improve the understanding of both clinicians and researchers.

Both the butterfly diagram and the hodograph provide unique methods for visualizing the gait parameters discussed in this thesis and reflect the impact of those parameters on the overall gait cycle. By integrating the use of these visualizations into clinical settings, new insights may be found from the expertise of physiotherapists and clinicians.

4.3.4. Estimated handrail usage by poststroke participants

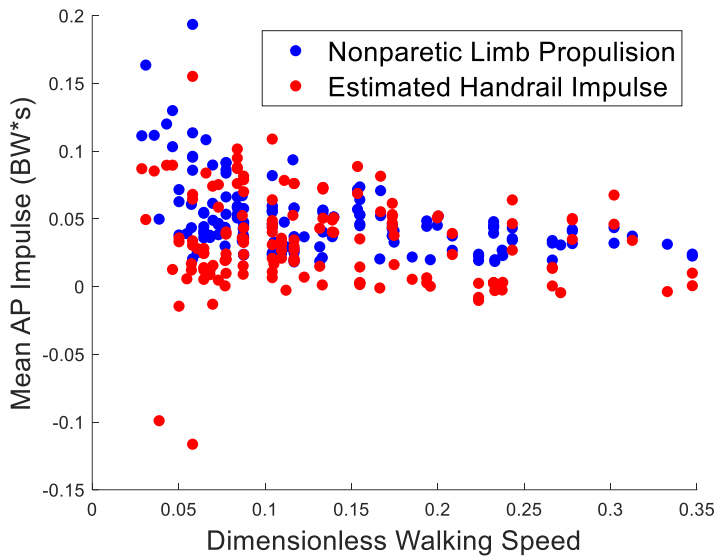

Figure 4-16: Estimated AP impulse attributed to handrail use by poststroke participants

Positive y-value indicates a propulsive impulse. Handrail impulse is estimated by the mean impulse value required to ensure periodic motion in the hodograph, returning to the same initial velocity at the end of the stride.

The usage of handrails by the poststroke participants is shown in Figure 4-16. The mean impulse for the handrail usage over a stride is similar to the magnitude of mean nonparetic limb propulsive impulse. When combined with the reduced impulse from the paretic limb, these values show that the participants are primarily augmenting propulsion with handrail use. The distribution of the force exerted on the handrails throughout the gait cycle is unknown, making estimation of the forces via this method not possible.

Chapter 5. Conclusion

5.1. Summary

This thesis has examined the quantitative parameters and visualizations that can be determined from the outputs of an instrumented treadmill in the context of subacute stroke rehabilitation. The normative values for neurologically intact gait at low walking speed, and the parameters' relationship to walking speed were identified. Changes in these relationships with walking speed were found in the subacute stroke population. Propulsive impulse, lateral impulse, propulsive force, CoM velocity, CoM acceleration, and CoP variability were all found to have altered in the subacute stroke population compared to controls.

A subset of poststroke participants was examined longitudinally over 1-2 months across all gait parameters. All of the poststroke participants who were followed longitudinally improved in the clinical gait measures after a 3-week follow-up assessment. Improvements in CoP variability, propulsion, step length asymmetry, and lateral CoM velocity range were seen over the course of rehabilitation. Within the subset, patterns potentially related to age and time post stroke were found in how recovery trajectories grouped together.

5.2. Strengths

This thesis provides kinetic analysis of subacute stroke which has not been used in previous literature. The range of variables investigated encompasses many of the methods used in chronic research, alongside methods only previously used in the analysis of amputee and neurologically intact gait.

This the first use of hodographs and butterfly diagrams for subacute stroke as far as the author is aware. This is also the first reported use of hodographs for very slow gait speeds, though it is possible to generate them using the smaller dataset provided in Wu (2019).

All data collection for subacute stroke was completed with minimal interference to ongoing rehabilitation. Overall setup for a subacute stroke participant takes less than 5

minutes on average, mainly attributed to harnessing and taking baseline blood pressure readings. Physiotherapists were present for all data collections, and a rapport was developed between researchers and PTs that will benefit future collaboration.

Additional metrics including velocity redirection angle were calculated, but they showed high sensitivity to low-speed values due to the method for determining them. Velocity difference across double support was created as a surrogate for this measure.

5.3. Limitations

Stroke participants were recommended to the pilot study by their physiotherapists, introducing a potential bias in recruitment towards less severe cases.

The analysis used 15 consecutive strides per trial to assess gait variability across the same number of strides for each participant. It is unknown if the true variability of certain metrics is captured by this number of strides. This gives the benefit of being an achievable amount of data per participant even after missteps are considered.

KDE for the CoP intersections uses an algorithm which is quite sensitive to parameters including the kernel size/number used. The sizes used in this thesis were based on the analysis used in previous studies, but this was sensitive to outliers. Further investigation of the impact of kernel size is warranted.

The use of handrails and body-weight support likely introduced a bias into the results of some individuals. Body-weight support has been shown to reduce requirements for lateral stability (Dragunas & Gordon, 2016). This effect was quantified and shown to be limited in total impact when offset by equivalent impulses. The timing of handrail usage and engagement with body-weight support was not quantified, so they exact point at which the deviations will take effect is something that requires further investigation.

5.4. Future directions

5.4.1. Improving metrics and system reliability

The instrumented treadmill system used in this study and future longitudinal analysis of subacute stroke gait could be improved. Having instrumented handrails and body-weight support would improve the quality of data in more severe cases. Instrumenting the bodyweight support could be accomplished by placing a transducer between the harness hanger and belt of the support device. Alternatively, it may be possible to measure the beam deflection of the overhead rails to which the bodyweight support is attached, the feasibility of this would depend on the vibration and dynamic loading response of the rails.

Instrumented the handrails has commercial solutions available from the manufacturer of the treadmill, but the feedback of the physiotherapists indicate that customizable handrails would be more useful. Balancing the need for instrumentation with adjustable handrail configurations will present future challenges. In the meantime, it could be beneficial to examine how handrail use impacts the vertical moments exerted and the lateral stability during walking. The role of upper limb swing in perturbation recovery and energetic cost has not been investigated at the slow walking speeds seen in this thesis. The use of handrails in clinical populations is debated, with some studies suggesting that it improves gait quality (Kang et al., 2014) while others indicate that it reduces motor learning ability (Buurke et al., 2019).

Adding self-pacing to the speed control of the treadmill could allow for investigation into a more natural gait. Allowing participants to adjust their walking speed dynamically would provide a better representation of overground walking. This however would introduce new complexity in characterizing gait at variable speeds within a session, requiring adjustments to calculations which depend on walking speed. In the hodograph this would require the integration constant for the forward velocity to be adjusted based on the current treadmill velocity or the mean velocity over the course of the stride. This analysis would

allow for a stronger investigation into how the shape of the hodograph changes with non steady walking speeds.

Implementing kinematic measures via markerless motion capture could allow for more in depth analysis of gait. Certain deficits such as spasticity, where joint range of motion is impacted, could be characterized more efficiently with added kinematics. The VR modules used in the system are able to use a markerless depth camera for added patient interactivity, but full 3D kinematics would require a multicamera system to be installed.

Investigation of the number of strides required to accurately assess gait variability is warranted for long term use of the system. The impact of non-consecutive strides should be assessed.

5.4.2. Subacute gait research

Future studies on the recovery of gait would benefit from an alteration to the test protocol. Having poststroke participants walk at the speeds used in previous sessions as well as their new comfortable speed would allow the assessment to detect what has changed to allow improvements. As more data is collected from subacute stroke patients, a more accurate characterization of recovery could be developed. Developing test protocols for each parameter will allow for more in depth investigation of their impact on overall gait.

The impact of sex and gender in stroke rehabilitation has been primarily investigated in terms of risk factors, access to care, and diagnosis (Ospel et al. 2023). The effect of sex and gender on gait metrics was not specifically investigated in this thesis, but the outputs of the treadmill could provide objective measures for such analysis. The treadmill would allow for investigation into potential biases present in the way poststroke gait is currently assessed. Prior to such investigations, it would be important for the potential bias in outcome measures from the treadmill to be evaluated. The known impacts of morphological differences between sexes on gait should be considered.

The pilot study which this data is apart of will examine the effectiveness of split-belt treadmill training. The parameters examined and developed in this thesis can be used to quantify the gait changes and compare the effectiveness of split-belt usage. Additional metrics to quantify gait adaptation can be derived from the parameters used here.

The effect of VR and biofeedback is another potential avenue of research that would benefit from the use of NeuroRecoVR and the parameters outlined in this thesis.

5.4.3. Knowledge translation for clinical use

NeuroRecoVR has the benefit of being designed with the feedback of physiotherapists. The data presented in chapter 4 was collected during scheduled rehabilitation sessions. In order for the physiotherapists to benefit from the use of quantitative treadmill data, future work should focus on the correlation between therapist observations and gait metrics. Knowing which metrics correlate to specific deficits would allow for tailored outcome measures for personalized rehabilitation. The next steps to achieve this goal would be to instate reporting or debriefing procedures to provide gait measures to the therapists. Meeting with clinical teams to review and evaluate the patient data would benefit both the therapists and researchers.

References

Adamcyzk, P. (2008). The Influence of Center of Mass Velocity Redirection on Mechanical and Metabolic Perfomance During Walking.

Adamczyk, P., & Kuo, A. (2009, 8). Redirection of center-of-mass velocity during the step-to-step transition of human walking. *Journal of Experimental Biology, 212*(16), 2668-2678.

Alexander, L., Black, S., Patterson, K., Gao, F., Danells, C., & Mcllroy, W. (2009, 2). Association between gait asymmetry and brain lesion location in stroke patients. *Stroke, 40*(2), 537-544.

Allen, J., Kautz, S., & Neptune, R. (2011, 4). Step length asymmetry is representative of compensatory mechanisms used in post-stroke hemiparetic walking. *Gait and Posture, 33*(4), 538-543.

Bauby, C., & Kuo, A. (2000). Active control of lateral balance in human walking. *Journal of Biomechanics, 33*, 1433-1440. Retrieved from http://www.elsevier.nl:80/

Bertram, J. (2005, 3). Constrained optimization in human walking: Cost minimization and gait plasticity. *Journal of Experimental Biology, 208*(6), 979-991.

Best, A., & Wu, A. (2020, 10). Upper body and ankle strategies compensate for reduced lateral stability at very slow walking speeds: Balance strategies at very slow speeds. *Proceedings of the Royal Society B: Biological Sciences, 287*(1936).

Beyaert, C., Vasa, R., & Frykberg, G. (2015). Gait post-stroke: Pathophysiology and rehabilitation strategies. *Neurophysiologie Clinique, 45*(4-5), 335-355.

Bonkhoff, A., Hope, T., Bzdok, D., Guggisberg, A., Hawe, R., Dukelow, S., . . . Bowman, H. (2022, 4). Recovery after stroke: the severely impaired are a distinct group. *Journal of neurology, neurosurgery, and psychiatry, 93*(4), 369-378.

Botev, Z., Grotowski, J., & Kroese, D. (2010, 10). Kernel density estimation via diffusion. *Annals of Statistics, 38*(5), 2916-2957.

Bowden, M., Balasubramanian, C., Neptune, R., & Kautz, S. (2006, 3). Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. *Stroke, 37*(3), 872-876.

Bruijn, S., Meijer, O., Beek, P., & Van Dieën, J. (2010, 12). The effects of arm swing on human gait stability. *Journal of Experimental Biology, 213*(23), 3945-3952.

Bruijn, S., van Dieën, J., Meijer, O., & Beek, P. (2009, 7). Is slow walking more stable? *Journal of Biomechanics, 42*(10), 1506-1512.

Buurke, T., Lamoth, C., Van Der Woude, L., & Den Otter, R. (2019, 9). Handrail holding during treadmill walking reduces locomotor learning in able-bodied persons. *IEEE Transactions on Neural Systems and Rehabilitation Engineering, 27*(9), 1753-1759.

Callegari, B., Garcez, D., da Cruz Júnior, A., Cardoso Almeida, A., Candeira, S., do Nascimento, N., . . . Costa e Silva, A. (2021, 12). Gait patterns in ischemic and hemorrhagic post-stroke patients with delayed access to physiotherapy. *Hong Kong Physiotherapy Journal, 41*(2), 77-87.

Cavagna, G. (1985). *Force platforms as ergometers.*

Collins, S., Adamczyk, P., & Kuo, A. (2009, 10). Dynamic arm swinging in human walking. *Proceedings of the Royal Society B: Biological Sciences, 276*(1673), 3679-3688.

Collins, S., Wisse, M., & Ruina, A. (2001). *A Three-Dimensional Passive-Dynamic Walking Robot with Two Legs and Knees.*

Croft, J., Schroeder, R., & Bertram, J. (2017, 12). The goal of locomotion: Separating the fundamental task from the mechanisms that accomplish it. *Psychonomic Bulletin and Review, 24*(6), 1675-1685.

Dean, J., Bowden, M., Kelly, A., & Kautz, S. (2020, 2). Altered post-stroke propulsion is related to paretic swing phase kinematics. *Clinical Biomechanics, 72*, 24-30.

Donelan, J., Kram, R., & Kuo, A. (2002). Simultaneous positive and negative external mechanical work in human walking. *Journal of Biomechanics, 35*, 117-124.

Donkor, E. (2018). Stroke in the 21st Century: A Snapshot of the Burden, Epidemiology, and Quality of Life. *Stroke Research and Treatment, 2018*.

Dragunas, A., & Gordon, K. (2016, 9). Body weight support impacts lateral stability during treadmill walking. *Journal of Biomechanics, 49*(13), 2662-2668.

Eng, J., & Tang, P. (2007, 10). Gait training strategies to optimize walking ability in people with stroke: A synthesis of the evidence. *Expert Review of Neurotherapeutics, 7*(10), 1417-1436.

Farris, D., Hampton, A., Lewek, M., & Sawicki, G. (2015, 2). Revisiting the mechanics and energetics of walking in individuals with chronic hemiparesis following stroke: From individual limbs to lower limb joints. *Journal of NeuroEngineering and Rehabilitation, 12*(1).

Feasel, J., Whitton, M., Kassler, L., Brooks, F., & Lewek, M. (2011, 6). The integrated virtual environment rehabilitation treadmill system. *IEEE Transactions on Neural Systems and Rehabilitation Engineering, 19*(3), 290-297.

Fellin, R., Rose, W., Royer, T., & Davis, I. (2010, 11). Comparison of methods for kinematic identification of footstrike and toe-off during overground and treadmill running. *Journal of Science and Medicine in Sport, 13*(6), 646-650.

Finley, J., Long, A., Bastian, A., & Torres-Oviedo, G. (2015, 9). Spatial and Temporal Control Contribute to Step Length Asymmetry during Split-Belt Adaptation and Hemiparetic Gait. *Neurorehabilitation and Neural Repair, 29*(8), 786-795.

Fugl-Meyer, A., Jaasko, L., Leyman, I., Olsson, S., & Steglind, S. (1975). The Post Stroke Hemiplegic Patient. *Scand J Rehab Med, 7*, 13-31.

Gard, S., Miff, S., & Kuo, A. (2004, 4). Comparison of kinematic and kinetic methods for computing the vertical motion of the body center of mass during walking. *Human Movement Science, 22*(6), 597-610.

Gelaw, A., Janakiraman, B., Teshome, A., & Ravichandran, H. (2019, 11). Effectiveness of treadmill assisted gait training in stroke survivors: A systematic review and meta-analysis. *Global Epidemiology, 1*.

Germanotta, M., Iacovelli, C., & Aprile, I. (2022, 10). Evaluation of Gait Smoothness in Patients with Stroke Undergoing Rehabilitation: Comparison between Two Metrics. *International Journal of Environmental Research and Public Health, 19*(20).

Goldberg, E., Kautz, S., & Neptune, R. (2008). Can treadmill walking be used to assess propulsion generation? *Journal of Biomechanics, 41*(8), 1805-1808.

Gowland, C., Stratford, P., Ward, M., Moreland, J., Torresin, W., Sandra, P., . . . Plews, N. (1993). Measuring Physical Impairment and Disability With the Chedoke-McMaster Stroke Assessment. *Stroke, 24*(1), 58-63. Retrieved from http://ahajournals.org

Guertin, P., & Steuer, I. (2009, 8). Key central pattern generators of the spinal cord. *Journal of Neuroscience Research, 87*(11), 2399-2405.

Harris, J., Eng, J., & Ot, P. (2004). Goal Priorities Identified through Client-Centred Measurement in Individuals with Chronic Stroke. *Physiotherapy Canada, 56*(3), 171-176.

Hawe, R., Scott, S., & Dukelow, S. (2019). Taking Proportional out of Stroke Recovery. *Stroke, 50*(1), 204-211.

Helm, E., & Reisman, D. (2015, 11). The Split-Belt Walking Paradigm. Exploring Motor Learning and Spatiotemporal Asymmetry Poststroke. *Physical Medicine and Rehabilitation Clinics of North America, 26*(4), 703-713.

Herzog, W., Nigg, B., Read, L., & Olsson, E. (1989). *Asymmetries in ground reaction force patterns in normal human gait* (Vol. 21).

Hsiao, H., Zabielski, T., Palmer, J., Higginson, J., & Binder-Macleod, S. (2016, 12). Evaluation of measurements of propulsion used to reflect changes in walking speed in individuals poststroke. *Journal of Biomechanics, 49*(16), 4107-4112.

Iyer, S., Joseph, J., & Vashista, V. (2020, 3). Evolving Toward Subject-Specific Gait Rehabilitation Through Single-Joint Resistive Force Interventions. *Frontiers in Neurorobotics, 14*.

Janeh, O., & Steinicke, F. (2021, 11). A Review of the Potential of Virtual Walking Techniques for Gait Rehabilitation. *Frontiers in Human Neuroscience, 15*.

Janeh, O., Fründt, O., Schönwald, B., Gulberti, A., Buhmann, C., Gerloff, C., . . . Pötter-Nerger, M. (2019, 5). Gait Training in Virtual Reality: Short-Term Effects of Different Virtual Manipulation Techniques in Parkinson's Disease. *Cells, 8*(5), 419.

Jang, S. (2010, 12). The recovery of walking in stroke patients: A review. *International Journal of Rehabilitation Research, 33*(4), 285-289.

Jones, P., Pomeroy, V., Wang, J., Schlaug, G., Tulasi Marrapu, S., Geva, S., . . . Baron, J. (2016, 2). Does stroke location predict walk speed response to gait rehabilitation? *Human Brain Mapping, 37*(2), 689-703.

Jorgensen, H., Nakayama, H., Raaschou, H., Olsen, T., & Jergensen, H. (1995). Recovery of Walking Function in Stroke Patients: The Copenhagen Stroke Study. *Arch Phys Med Rehabil, 76*.

Kalron, A., & Frid, L. (2015, 11). The "butterfly diagram": A gait marker for neurological and cerebellar impairment in people with multiple sclerosis. *Journal of the Neurological Sciences, 358*(1-2), 92-100.

Kalron, A., Dvir, Z., Frid, L., & Achiron, A. (2013, 6). Quantifying Gait Impairment Using an Instrumented Treadmill in People with Multiple Sclerosis. *ISRN Neurology, 2013*, 1-6.

Kang, K., Lee, K., Son, S., Kwon, J., & Kim, K. (2015). Effect of handrail use while performing treadmill walking on the gait of stroke patients. *Journal of Physical Therapy Science*, 833-835.

Kanko, R., Laende, E., Selbie, W., & Deluzio, K. (2021, 5). Inter-session repeatability of markerless motion capture gait kinematics. *Journal of Biomechanics, 121*.

Klassen, T., Dukelow, S., Bayley, M., Benavente, O., Hill, M., Krassioukov, A., . . . Eng, J. (2020, 9). Higher Doses Improve Walking Recovery during Stroke Inpatient Rehabilitation. *Stroke*, 2639-2648.

Kribus-Shmiel, L., Zeilig, G., Sokolovski, B., & Plotnik, M. (2018, 2). How many strides are required for a reliable estimation of temporal gait parameters? Implementation of a new algorithm on the phase coordination index. *PLoS ONE, 13*(2).

Kroneberg, D., Elshehabi, M., Meyer, A., Otte, K., Doss, S., Paul, F., . . . Schmitz-Hübsch, T. (2019). Less is more - Estimation of the number of strides required to assess gait variability in spatially confined settings. *Frontiers in Aging Neuroscience, 11*(JAN).

Kuo, A., & Donelan, J. (2010). Dynamic Principles of Gait and Their Clinical Implications. *Physical Therapy, 90*(2), 157. Retrieved from www.ptjournal.org

Lee, S., & Hidler, J. (2008). Biomechanics of overground vs. treadmill walking in healthy individuals. *J Appl Physiol, 104*, 747-755. Retrieved from http://www.

Lee, Y., & Liang, J. (2020, 2). Characterizing intersection variability of butterfly diagram in post-stroke gait using Kernel Density Estimation. *Gait and Posture, 76*, 157-161.

Levin, I., Lewek, M., Feasel, J., & Thorpe, D. (2017). Gait training with visual feedback and proprioceptive input to reduce gait asymmetry in adults with cerebral palsy: A case series. *Pediatric Physical Therapy, 29*(2), 138-145.

Lewek, M. (2011, 1). The influence of body weight support on ankle mechanics during treadmill walking. *Journal of Biomechanics, 44*(1), 128-133.

Lewek, M., Raiti, C., & Doty, A. (2018, 12). The Presence of a Paretic Propulsion Reserve During Gait in Individuals Following Stroke. *Neurorehabilitation and Neural Repair, 32*(12), 1011-1019.

Li, S., Francisco, G., & Zhou, P. (2018, 8). Post-stroke hemiplegic gait: New perspective and insights. *Frontiers in Physiology, 9*(AUG).

Mahon, C., Farris, D., Sawicki, G., & Lewek, M. (2015, 4). Individual limb mechanical analysis of gait following stroke. *Journal of Biomechanics, 48*(6), 984-989.

Meurisse, G., Bastien, G., & Schepens, B. (2019, 8). The step-to-step transition mode: A potential indicator of first-fall risk in elderly adults? *PLoS ONE, 14*(8).

Meurisse, G., Bastien, G., & Schepens, B. (2019a, 1). Effect of age and speed on the step-to-step transition phase during walking. *Journal of Biomechanics, 83*, 253-259.

Meyer, C., Killeen, T., Easthope, C., Curt, A., Bolliger, M., Linnebank, M., . . . Filli, L. (2019, 12). Familiarization with treadmill walking: How much is enough? *Scientific Reports, 9*(1).

Mohan, D., Khandoker, A., Wasti, S., Ismail Ibrahim Ismail Alali, S., Jelinek, H., & Khalaf, K. (2021, 6). Assessment Methods of Post-stroke Gait: A Scoping Review of Technology-Driven Approaches to Gait Characterization and Analysis. *Frontiers in Neurology, 12*.

Moore, J. K., Hnat, S. K., & van den Bogert, A. J. (2015). An elaborate data set on human gait and the effect of mechanical perturbations. *PeerJ, 3*, 918.

Nakano, N., Sakura, T., Ueda, K., Omura, L., Kimura, A., Iino, Y., . . . Yoshioka, S. (2020, 5). Evaluation of 3D Markerless Motion Capture Accuracy Using OpenPose With Multiple Video Cameras. *Frontiers in Sports and Active Living, 2*.

Nascimento, L., Boening, A., Galli, A., Polese, J., & Ada, L. (2021, 4). Treadmill walking improves walking speed and distance in ambulatory people after stroke and is not inferior to overground walking: a systematic review. *Journal of Physiotherapy, 67*(2), 95-104.

Osoba, M., Rao, A., Agrawal, S., & Lalwani, A. (2019, 2). Balance and gait in the elderly: A contemporary review. *Laryngoscope Investigative Otolaryngology, 4*(1), 143-153.

Ospel, J., Singh, N., Ganesh, A., & Goyal, M. (2023, 1). Sex and Gender Differences in Stroke and Their Practical Implications in Acute Care. *Journal of Stroke, 25*(1), 16-25.

Owings, T., & Grabiner, M. (2003, 8). Measuring step kinematic variability on an instrumented treadmill: How many steps are enough? *Journal of Biomechanics, 36*(8), 1215-1218.

Owings, T., & Grabiner, M. (2004, 6). Step width variability, but not step length variability or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion. *Journal of Biomechanics, 37*(6), 935-938.

Penke, K., Scott, K., Sinskey, Y., & Lewek, M. (2019, 6). Propulsive Forces Applied to the Body's Center of Mass Affect Metabolic Energetics Poststroke. *Archives of Physical Medicine and Rehabilitation, 100*(6), 1068-1075.

Plotnik, M., Azrad, T., Bondi, M., Bahat, Y., Gimmon, Y., Zeilig, G., . . . Siev-Ner, I. (2015). Self-selected gait speed - Over ground versus self-paced treadmill walking, a solution for a paradox. *Journal of NeuroEngineering and Rehabilitation, 12*(1).

Reed, L., Urry, S., & Wearing, S. (2013). Reliability of spatiotemporal and kinetic gait parameters determined by a new instrumented treadmill system. *BMC Musculoskeletal Disorders, 14*.

Reisman, D., Wityk, R., Silver, K., & Bastian, A. (2007, 7). Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke. *Brain, 130*(7), 1861-1872.

Richards, C., Malouin, F., Lamontagne, A., McFadyen, B., Dumas, F., Comeau, F., . . . Fung, J. (2018, 6). Gait training after stroke on a self-paced treadmill with and without virtual environment scenarios: A proof-of-principle study. *Physiotherapy Canada, 70*(3), 221-230.

Rouhani, H., Abe, M., Nakazawa, K., Popovic, M., & Masani, K. (2015, 3). Heel strike detection using split force-plate treadmill. *Gait and Posture, 41*(3), 863-866.

Rowland, D., & Lewek, M. (2022, 9). Linking gait mechanics with perceived quality of life and participation after stroke. *PLoS ONE, 17*(9 September).

Selves, C., Stoquart, G., & Lejeune, T. (2020, 8). Gait rehabilitation after stroke: review of the evidence of predictors, clinical outcomes and timing for interventions. *Acta Neurologica Belgica, 120*(4), 783-790.

Sheffler, L., & Chae, J. (2015, 11). Hemiparetic Gait. *Physical Medicine and Rehabilitation Clinics of North America, 26*(4), 611-623.

Sloot, L., Houdijk, H., & Harlaar, J. (2015, 6). A comprehensive protocol to test instrumented treadmills. *Medical Engineering and Physics, 37*(6), 610-616.

Sloot, L., van der Krogt, M., & Harlaar, J. (2014). Effects of adding a virtual reality environment to different modes of treadmill walking. *Gait and Posture, 39*(3), 939-945.

Sloot, L., van der Krogt, M., & Harlaar, J. (2014, 1). Self-paced versus fixed speed treadmill walking. *Gait and Posture, 39*(1), 478-484.

Smith, A., Lemaire, E., & Nantel, J. (2018, 9). Lower limb sagittal kinematic and kinetic modeling of very slow walking for gait trajectory scaling. *PLoS ONE, 13*(9).

Soo, C., & Donelan, J. (2012, 2). Coordination of push-off and collision determine the mechanical work of step-to-step transitions when isolated from human walking. *Gait and Posture, 35*(2), 292-297.

Takakusaki, K. (2017). Functional Neuroanatomy for Posture and Gait Control. Retrieved from www.e-jmd.org

Teasell, R., Bitensky, J., Salter, K., & Bayona, N. (2005, 6). The role of timing and intensity of rehabilitation therapies. *Topics in Stroke Rehabilitation, 12*(3), 46-57.

Teasell, R., Hussein, N., Mrehabmed, M., & Longval Bsc, M. (2020). *EBRSR Clinicians Handbook.* Retrieved from www.ebrsr.com

Tesio, L., Rota, V., & Perucca, L. (2011, 2). The 3D trajectory of the body centre of mass during adult human walking: Evidence for a speed-curvature power law. *Journal of Biomechanics, 44*(4), 732-740.

Tesio, L., Scarano, S., Cerina, V., Malloggi, C., & Catino, L. (2021, 6). Velocity of the Body Center of Mass during Walking on Split-Belt Treadmill. *American Journal of Physical Medicine and Rehabilitation, 100*(6), 620-624.

Thomas, S., Vega, D., & Arellano, C. (2021, 1). Do humans exploit the metabolic and mechanical benefits of arm swing across slow to fast walking speeds? *Journal of Biomechanics, 115*.

Urbaniak, G. C., & Plous, S. (2013). *Research Randomizer*. Retrieved from http://www.randomizer.org/

Usherwood, J. (2010, 1). Inverted pendular running: a novel gait predicted by computer optimization is found between walk and run in birds. *Biology Letters, 6*, 765-768.

Wade, L., Needham, L., McGuigan, P., & Bilzon, J. (2022, 2). *Applications and limitations of current markerless motion capture methods for clinical gait biomechanics* (Vol. 10). PeerJ Inc.

Whittle, M. (1997). Three-dimensional motion of the center of the body during walking. *Human Movement Science, 16*, 347-355.

Wonsetler, E., & Bowden, M. (2017). A systematic review of mechanisms of gait speed change post-stroke. part 1: Spatiotemporal parameters and asymmetry ratios. *Topics in Stroke Rehabilitation, 24*(6), 435-446.

Wonsetler, E., & Bowden, M. (2017). A systematic review of mechanisms of gait speed change Post-Stroke. Part 2: Exercise capacity, muscle activation, kinetics, and kinematics. *Topics in Stroke Rehabilitation, 24*(5), 394-403.

Wu, A., Simpson, C., van Asseldonk, E., van der Kooij, H., & Ijspeert, A. (2019, 12). Mechanics of very slow human walking. *Scientific Reports, 9*(1).

Zeni, J., Richards, J., & Higginson, J. (2008). Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait Posture*, 710-714.

Appendix A: Participant demographics

Table A-1: Neurologically intact participant demographics

ID	Age	Sex	Mass (kg)	Self Selected Walking Speed (m/s)
RVUG	27	M	111.5	1.64
JXTE	19	M	94.7	1.25
UVWR	30	F	55.1	1.18
WLPQ	23	F	65.1	1.04
GWGO	30	F	64	1.95
XOAC	35	M	83.7	1.25
DTGS	22	F	52.2	1.29
LKEG	75	M	84.8	1.31
LVVT	29	F	79.0	1.38
SOVA	57	F	92.1	*
JDKA	18	F	45.4	1.5
IYOW	22	F	82.7	1.43
GNDR	29	F	73.4	1.42
RCMD	60	M	88.7	1.29

*Self selected speed not collected due to time constraint

Table A-2: Poststroke participant demographics

ID	Age	Sex	Mass (kg)	Days Since Stroke at first assessment	Affected Side	FM-LE score (L - R)	10MWT	6MWT
YJSHD	58	M	78.2	42	Left	31-34	1.148	340
VHJQQ	63	M	72.8	189	Left	21-34	0.41	85
GGOOV	51	F	60.6	54	Right	34-22 34-27	0.476 0.90	136 280
BUQBS	51	F	110	73	Right	34-19 34-25	0.23 0.17#	84,98
FISCT	49	M	66.7	46	Left	26-34	0.96	258
HYBNV	43	F	58	39	Right	34-20	0.68	225
QYVAT	31	M	77.5	*	Left	31-34	1.22	415
GAQPQ	75	F	53	174	Left	15-34 15-34	0.18 0.23	55 66
VUAMB	48	F	138.4	139	Left	27-34	0.849	272
FXVQU	74	F	61.5	45	Left	25-34	0.455	149
QNASO	53	M	110.2	125	Right	*	*	*
CUEGK	83	M	85.6	16	Right	34-19 34-26	0.46 0.64	137 189
BOFTP	64	F	67.9	47	Left	17-34	0.04	0
DQPFO	45	F	62.2	30	Left	18-34	0.17	0

*Participants with missing information or lacking assessments were included in the group comparisons but not used for recovery metrics vs time or clinical measures, Fugl Meyer Lower extremity (FM-LE) has maximum score of 34

Participant moved from using a 4-wheel walker to a quad cane, reducing the result of the 10MWT