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Development and Application of 3D Kinematic Methodologies for Biomechanical Modelling in Adaptive Sports and Rehabilitation

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DEVELOPMENT AND APPLICATION OF 3D KINEMATIC METHODOLOGIES FOR BIOMECHANICAL MODELLING IN ADAPTIVE SPORTS AND REHABILITATION

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by
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ABSTRACT

Biomechanical analysis is widely used to assess human movement sciences, specifically using three-dimensional motion capture modelling. There are unprecedented opportunities to increase quantitative knowledge of rehabilitation and recreation for disadvantaged population groups. Specifically, 3D models and movement profiles for human gait analysis were generated with emphasis on post-stroke patients, with direct model translation to analyze equivalent measurements while horseback riding in use of the alternative form of rehabilitation, equine assisted activities and therapies (EAAT) or hippotherapy (HPOT). Significant improvements in gait symmetry and velocity were found within an inpatient rehabilitation setting for patients following a stroke, and the developed movement profiles for patients have the potential to address patient recovery timelines. For population groups, such as those following a cerebral incident, alternative forms of rehabilitation like EAAT and HPOT are largely unexplored. Within these studies, relevant muscular activations were found between healthy human gait and horseback riding, supporting the belief that horseback riding can stimulate similar movements within the rider. Even more, there was a strong correlation between the horse’s pelvic rotations, and the responsive joint moments and rotations of the rider. These findings could have greater implications in choosing horses, depending on the desired physical outcome, for EAAT and HPOT purposes. Similar approaches were also used to address another biomechanically disadvantage population, adaptive sport athletes. Utilizing similar methodologies, a novel 3D wheelchair tennis athlete model was created to analyze match-simulation assessments. Significant findings were present in the energy

expenditure between two drill assessments. Overall, the quantitative results, coupled with the qualitative assessment chapter, provide a robust assessment of the effects of 3D movement analysis on rehabilitation and adaptive activities.

DEDICATION

I am grateful to all of my mentors, colleagues, friends, and family members for inspiring and motivating my journey of this dissertation. Thereby, I dedicate my dissertation work to my husband, Lane, for being my constant support system. He never failed to encourage me and ensure I gave all of my effort to my work. I also dedicate this work my parents and sister, for always reminding me of my potential and their inspiration in my endeavors since I was a young girl. Thank you for being examples of dedication and hard work. My love for all of you can never be quantified.

ACKNOWLEDGEMENTS

Throughout my years at Clemson University, I was grateful to have received a great deal of support and assistance. First, I would like to thank my supervisor Dr. John DesJardins, who's guidance was invaluable while navigating my graduate endeavors. His constant excitement to delve into new research questions was enlightening as a young researcher, and I am grateful for everything I learned under your leadership. More, I would like to acknowledge my other committee members Dr. Kristine Vernon, Dr. Marieke Van Puymbroeck, and Dr. Richard Blob for their expertise and their willingness to sharpen my knowledge in each of their separate fields.

I would also like to thank Prisma Health for funding our pilot project to assess human gait performance for post-stroke patients during their inpatient recovery. This funding made it possible to achieve actual rehabilitation data within the hospital setting. Additionally, I would also like to acknowledge out funding from the Brooks Sports Science Institute and their support in our research endeavors with the Clemson Wheelchair Tennis team.

Even more, I'd like to acknowledge my co-authors that assisted in the publication, or preparation for publication, of my research endeavors. The qualitative assessment of Chapter Three would not have been present without Dr. Marieke Van Puymbroeck and her expertise. Additionally, my undergraduate researcher Nathan Luzum was vital in the data collection and co-writing of Chapter Four. I am also truly grateful for my other undergraduate researchers whose dedication was valuable for completion of my work.

Finally, I'd like to acknowledge the Bioengineering Department, my colleagues, and the programs I was involved in during my time at Clemson University. I was inspired and shaped into a well-rounded researcher and human being thanks to everyone's support and encouragement.

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CHAPTER ONE

QUANTIFICATION OF POST-STROKE INPATIENT REHABILITATION BIOMECHANICAL OUTCOME MEASUREMENTS

Stroke is one of the leading health care problems across the globe. It is widely accepted that physical rehabilitation after stroke should begin immediately in order to capture the spontaneous recovery phase, and that therapists should place a heavy focus on pelvic and core mobility methods to assist in recovery of gait biomechanics. However, very little is quantified regarding the biomechanics of acute post-stroke recovery and the effect of pelvic-core mobility biomechanics on clinical outcomes. This chapter describes the development and quantification of post-stroke inpatient rehabilitation measures for a pilot group of patients.

INTRODUCTION

Mobility assessment in stroke rehabilitation and its impact in healthcare

Stroke is a significant health care problem, with over 795,000 incidences in the USA and 15 million incidences worldwide each year (Mozaffarian et al., 2016; WHO's Global Health Estimates, n.d.). It is the third leading cause of death in the USA and the second leading cause of disability worldwide. Approximately 70-80% of stroke patients suffer from initial gait impairment from hemiparesis or partial paralysis (Thaut, McIntosh, & Rice, 1997). Post stroke physical rehabilitation has the ultimate goal of returning patients to their physical Activities of Daily Living (ADLs) (Hebert et al., 2016). Unfortunately, physical recovery outcomes are widely disparate, and the diversity of initial impairment, range of patient therapeutic response and wide potential of outcomes, often results in highly subjective and disparate physical rehabilitation practices. This makes evidence based physical rehabilitation outcomes difficult to assess (Kwakkel, Kollen, & Lindeman, 2004).

Importance of pelvic-core mobility in rehabilitation

Gait abnormalities from initial stroke impairment include nonuniform stride length, abnormal muscle activation, and lower extremity velocity (Thaut et al., 1997). These impairments stem from limited pelvic and lower limb mobility and is a top rehabilitation priority following stroke (Kong, Jeong, & Kim, n.d.). A large focus of gait correction in stroke patients is concentrated on pelvic tilt exercises to increase postural stability and trunk control that is essential for gait recovery (Kong et al., n.d.). A study found that patients who were admitted with categorized “severe limitation in mobility” and discharged with “less limitation in mobility” spent 32.9% of their Physical Therapy (PT) time on gait rehabilitation (Latham et al., 2005). Pelvic-core and lower limb training was a specific focus of this work, but biomechanically these pelvic core rehabilitation practices have not been well quantified. The effect of enhanced pelvic-core rehabilitation (e.g. range of motion, frequency, intensity and duration) on clinical outcomes is poorly understood. The biomechanical quantification of enhanced pelvic-core therapeutic practices could provide evidence-based measures and therapeutic protocol outcomes for acute stroke recovery.

Current biomechanical treatment approaches for stroke rehabilitation

Questions remain regarding the efficacy of rehabilitation task intensity, complexity, and timing in stroke rehabilitation (Langhorne, Bernhardt, & Kwakkel, 2011;

Richards, Malouin, & Nadeau, 2015). However, there is general consensus that increased physical intensity of interventions positively influences clinical outcomes, both in task specific and traditional therapies (Langhorne et al., 2011). This increased biomechanical intensity can take the form of more intensive training, more frequent training, or a combination of both. It has also been shown that aerobic exercises, combined with skilled training, have even more promising results (Hasan, Rancourt, Austin, & Ploughman, 2016). Stroke rehabilitation can begin within 48 hours of the onset if the patient is physiologically stable (Dobkin, 2004; Kwakkel et al., 2004; Langhorne et al., 2011), with the first five weeks being the most vital to the rate of recovery. Overall, further quantification of the biomechanical outcomes during treatment, specifically in the assessment of movement intervention and outcomes are needed to better understand the relationship between gait focused-rehabilitation interventions and early physical recovery.

RESEARCH METHODS

Study design and patient recruitment

Three subjects participating in this study were recruited through the Roger C. Peace Rehabilitation Hospital of the Greenville Health System. The study was approved by the Institutional Review Board (IRB) prior to any Human Subject Research activities. All stroke patients seen by Roger C. Peace therapies were carefully screened for study participation. Patients meeting study inclusion/exclusion criteria were approached by an IRB approved study investigator and asked to participate in the study.

Following IRB approval, the clinical partners at Roger C. Peace Rehabilitation Hospital at Greenville Health System lead the recruitment of subjects for this study. The subjects recruited were required to meet the following inclusion criteria: i) ages 30-75 years old, ii) considered a moderate stroke by clinical measures, and iii) ability to walk 3 meters with moderate assistance. Exclusion criteria included: i) previous history of neurological diseases, ii) any orthopedic or rheumatic condition that could interfere with data quality, iii) known vestibular dysfunction, and iv) non-correctable visual disabilities.

Baseline evaluations were collected during Session 1, immediately following patient clearance for participation. Evaluations continued every 2-3 days (Session 2, 3, etc.) during each participant's inpatient stay. Additional details will be described in following sections. Physical therapists or support staff assigned by hospital administration were present during the evaluations to ensure safety or assist in the three-meter walk of the patients. Study team members prepared the equipment and collected the kinematic and EMG data during the evaluation. The patient's physical therapist administered their activity level clinical assessment. The evaluations were conducted prior to the start of their physical therapy session.

Standardization of therapy

Study participation was anticipated to have little effect on the patient's standard physical therapy. All enrolled patients remained under standard of care therapy sessions based on the treating therapists' clinical decisions.

Kinematic model development

Preliminary work funded (Fall 2020) by an internal seed-grant with a clinical collaborator used 3D motion capture and EMG systems to collect initial data on quantifying standard post-stroke rehabilitation practices, to provide quantitative, evidence based metrics to assess outcome gains. A visual model to track human kinematic and EMG gait patterns was generated. Within the model generation software, the upper body of the gait model was built in parallel to the RAB upper extremity model (Rab, Petuskey, & Bagley, 2002), with a modification of the skull marker being in the anterior location (forehead) instead of the most superior position. The pelvis was built with right and left Iliac Crest and Greater Trochanter markers to define the Visual 3D pelvis. The relation of these markers yielded pelvic tilt. The lower limbs were also built based on marker placements supported by literature (C-Motion, n.d.). Finally, to fully capture the gait profile, force plates (AMTI Force and Motion, Watertown, MA) were integrated to the motion capture volume to track the stride events of Left Toe Off (LTO), Right Toe Off (RTO), Left Heel Strike (LHS), and Right Heel Strike (RHS).

Table 1. Lower body segment geometry and defining points to generate human gait model. Right and left limbs were labelled as "R" and "L" abbreviations.

Name	Shape	Location	Defining Point 1	Defining Point 2
Pelvis	Cylinder	Proximal	RIAS	LIAS
		Distal	RIPS	LIPS
R/L Thigh	Cylinder	Proximal	RIGHT_HIP/LEFT_HIP	Radius: 0089 m
		Distal	RFLE/LFLE	Radius: 0.5*R/LKNEEWIDTH+MARKER_RADIUS
R/L Shank	Cylinder	Proximal	RFLE/LFLE	RIGHT_KNEE
		Distal	RFAL	Radius: 0.5*R/LANKLEWIDTH+MARKER_RADIUS
R/L Foot	Cone	Proximal	RFAL/LFAL	RIGHT_ANKLE/LEFT_ANKLE
		Distal	R5M	R2M

Table 2. Upper body segment geometry and defining points to generate human gait model. Right and left limbs were labelled as “R” and “L” abbreviations.

Name	Shape	Location	Defining Point 1	Defining Point 2
Head	Ellipsoid	Proximal	T2	Radius: 0.01 m
		Distal	R_EAR	L_EAR
R/L Upper Arm	Cylinder	Proximal	R/LSHJC_STATIC	Radius: 0.17*Distance(LSHLDR,RSHLDR)
		Distal	R/LEJC_STATIC	Radius: 0.13*Distance(R/LELB,R/LDU)
		Extra Target	Posterior, R/LELB	-
R/L Hand	Cylinder	Proximal	Lateral: (R/L)DR	Medial: (R/L)DU
		Distal	Joint Center: (R/L)MC3	Radius: 0.5*distance((R/L)DR,(R/L)DU)
R/L Forearm	Cylinder	Proximal	R/LEJC_STATIC	Radius: R/LAR_DISTAL_RADIUS
		Distal	RDR/LDR	RDU/LDU
R/L Hand	Sphere	Proximal	R/LWJC_STATIC	Radius: 0.5*Distance(R/LDR,R/LDU)
		Distal	R/LHANDJC_STATIC	Radius: 0.5*Distance(R/LDR,R/LDU)
		Extra Target	Lateral, RDR/LDR	-

Kinematic Methods

Post cerebral accident and once cleared for rehabilitation therapies, subjects performed their Functional Abilities and Goals assessment with their PT for a baseline, quantitative, clinical measurement. During their last inpatient session, the subjects were given the activity level assessment again to compare initial spontaneous recovery phase outcomes. These measurements were reliable tests for motor impairment outcomes for stroke patients, and were already used consistently within the inpatient rehabilitation protocols.

In addition, physical gait function was assessed using 3-dimensional motion capture analysis (Qualisys, Goteborg, Sweden). This was a portable, six infra-red camera, motion capture system with software capabilities to capture desired joint, pelvis motion, and gait analysis by placing small reflective markers on specific anatomical landmarks.

During the assessments, the reflective markers captured the motion and coordination of the subject's pelvis and other desired joints at a walk. Gait assessments were conducted with a three-meter walkway. Eight cameras surrounded the volume. The participant was suited with pre-determined, anatomically positioned, reflective markers. Initially, the participant was asked to remain standing within the center of the volume for a static measurement. Then, at their own pace, each participant walked through the volume three-five times. Specifically, the software will assess three-dimensional roll (obliquity), pitch (tilt), and yaw (rotation) movements and standard gait assessments. The patients will be assessed on the symmetry and improvements of the following:

Table 3. Kinematic variable and gait metrics used to assess patient progress in rehabilitation. Visual 3D was used as the primary method for data analysis and report generation. All variables were collected in metric units.

Parameter	Method	Method Units
Pelvic tilt (α), obliquity (β), and rotation (ϕ)	Visual 3D analysis	Degrees (°)
Stride length (L)	Visual 3D analysis	Meters (m)
Gait velocity (v)	Visual 3D analysis	Meters/Seconds (m/s)
Hip flexion/extension (σ)	Visual 3D annalysis	Degrees (°)

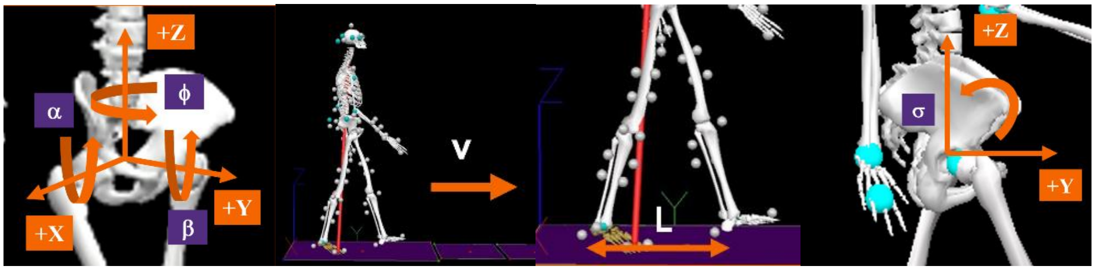

Figure 1. Visual representation of kinematic variables and gait metrics within Visual 3D.

The motion capture data was normalized (0-100% of stride) and filtered for any outliers in the data. All data is exported from Qualisys to Visual3D (C-Motion Research Biomechanics, MD) for analysis. One stride is taken from each pass, from left heel strike to the subsequent left heel strike. A custom pipeline analysis of data and reports were generated to graphically represent desired kinematic and gait metric values. An Average Filter was used to smooth kinematic data and remove any outliers. Averages of the desired calculations were taken from three-five passes and data normalized to 0-100% of the gait cycle. For kinematic data, total ranges of motion were calculated from average waveforms of joint movement during the gait cycle. Range of motion was compared between data sets using a two-sample T-test, assuming equal variances. Similarly, gait metrics were analyzed using two-sample T-test, assuming equal variances.

EMG Methods

Wireless EMG (Delsys) sensors were attached to the subjects and provided a real-time measure of the subject's muscle activation patterns during gait tracking. They were placed on the patient's trunk and legs to track muscular contraction in the right and left (1) rectus femoris, (2) Iliopsoas, (3) thoracolumbar fascia (4) rectus abdominis, and (5) biceps femoris (Figure 2). While other muscle groups are pertinent towards stroke recovery, their excitation was not be tracked in this study because muscle groups were chosen based on ones comparable to horseback riding interventions that are the focus of other aims in this dissertation. The sensors were placed according to SENIAM guidelines (Stegeman & Hermens, n.d.) and prior studies (Cuesta-Vargas & González-Sánchez, 2013). These sensors integrated with the motion capture technologies to simultaneously track muscular activation with corresponding gait events. All measurements sets were conducted *prior* to the start of their PT session for consistency in human performance. Overall, with a 3D motion capture volume integrated with a force plate walkway and

EMG tracking, robustly captured the spontaneous recovery phase during inpatient rehabilitation.

To process the EMG data, a custom analysis pipeline and report generation was also used to rectify and filter the raw EMG data (Bandpass, 50 - 500 Hz), as well as to normalize all data points as a percentage of the maximum contraction. After this, the data was normalized to 0 - 100% of the gait cycle, with an averaged data point at each percentage point (i.e. 101 data points for 0 - 100%). This data was averaged for each of the three-five strides to create one characteristic pattern for each muscle.

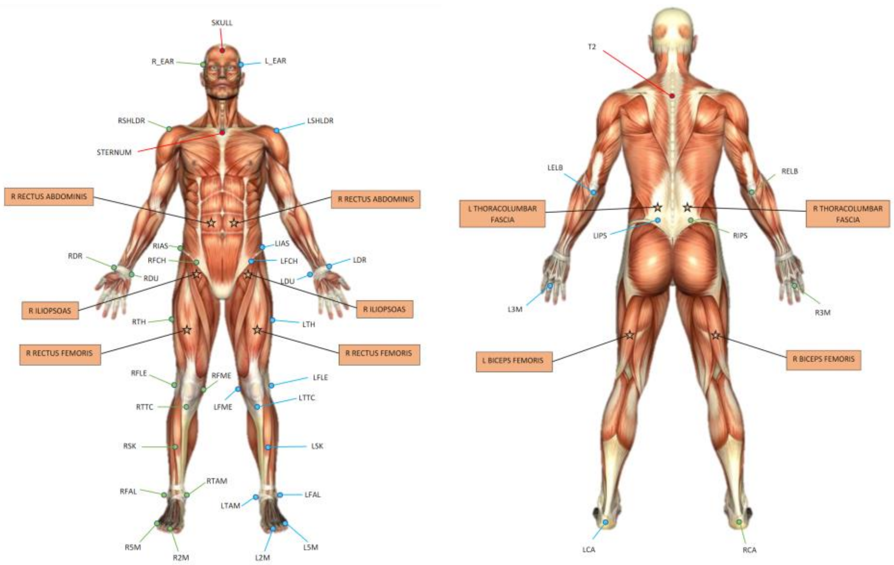

Figure 2. Anatomical locations for reflective markers to generate human segments and post analyse the gait model. EMG sensor locations for muscular activity tracking during data collection.

RESULTS

Patient 1 (P1)

The first patient was a Caucasian male, 32 years old, and suffered a mild/moderate stroke. Two data sets were collected for P1, data set 1 (DS1) and data set 2 (DS2) on Day 1 and Day 7, respectively. Visually, results for pelvic tilt, obliquity, and rotation show an increase in range of motion in pelvic tilt and rotation (Figure 3). This is also evident by the calculated values between DS1 and DS2, though no significant increase in motion were found (Table 4). More, the range of motion in the left hip flexion/extension significantly improved from DS1 to DS2, and was found to be significantly smaller than right hip movement within DS2. However, within DS2, there was no significant difference in flexion/extension between right and left hip movements (Table 4).

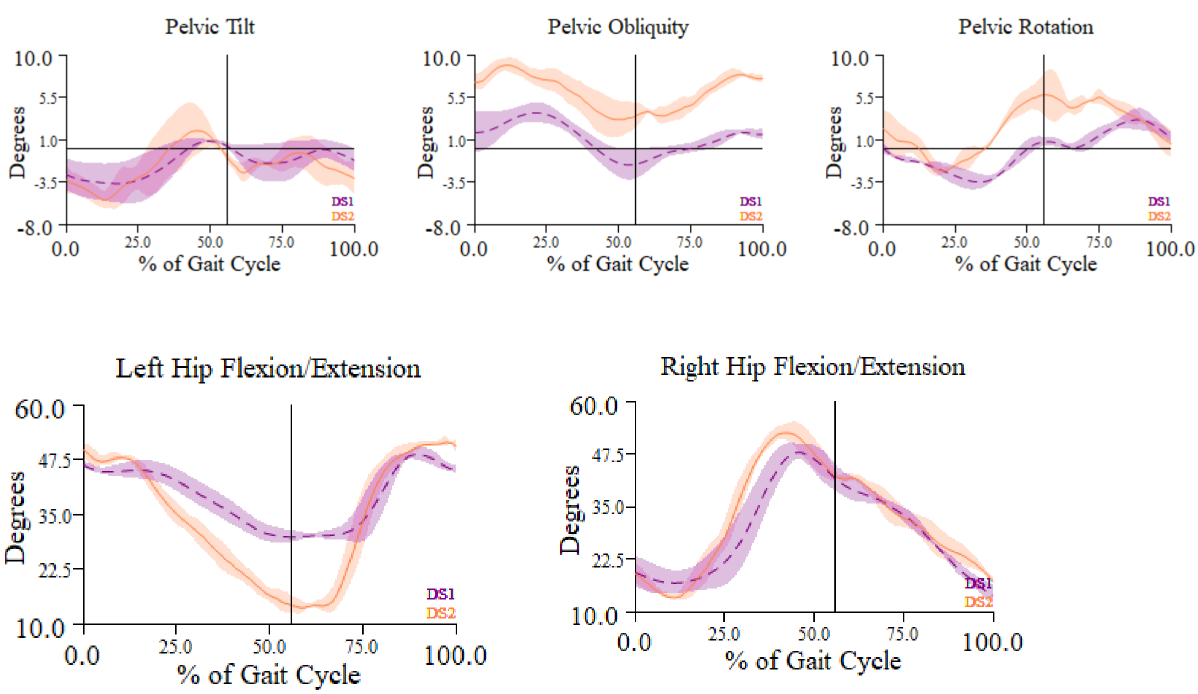

Figure 3. Kinematic waveforms of P1 pelvic tilt (top left), pelvic obliquity (top middle), pelvic rotation (top right), left hip flexion (+) and extension (-) (bottom left), and right hip flexion (+) and extension (-) (bottom right) during their gait stride, for DS1 (purple, dashed) and DS2 (orange, solid).

Table 4. Quantitative values for range of motion in the desired kinematic assessments. Statistically significant differences were present between DS1 and DS2 (*) and between right and left measurements (**).

Data Set	α (°)	β (°)	ϕ (°)	Left Hip σ (°)	Right Hip σ (°)
DS1	4.78±1.88	5.55±1.86	6.90±1.70	20.23±1.39*,**	34.73±2.88**
DS2	6.73±1.84	6.00±0.29	7.76±0.87	36.96±0.73*	36.97±1.62

Gait metrics, stride length and stride time, were divided between left and right steps. Between DS1 and DS2, left step lengths significantly increased and, consequently, right step length significantly decreased. As a result, more even gait steps were seen between DS1 and DS2, however, there was still a significant difference between right and left step lengths during both data sets (Figure 4). Similarly, a significant difference in step time was found between the right and left steps for DS1 and DS2 data sets. However, no significant improvements in step time for either right or left steps were found (Figure 5). Overall gait velocity was found to significantly improve between DS1 and DS2 gait cycles (Figure 6).

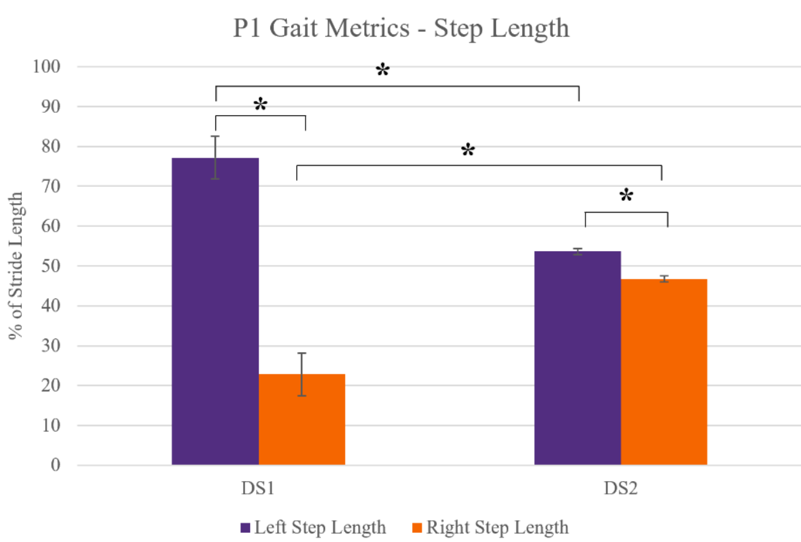

Figure 4. Gait step lengths for DS1 and DS2 normalized from 0-100% of total stride length. Statistically significant differences data sets and gait symmetry represented by *.

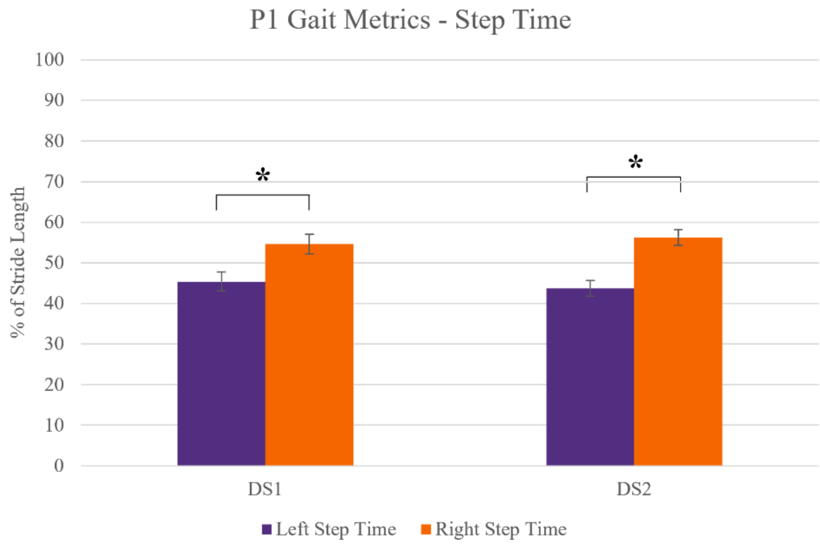

Figure 5. Gait step time for DS1 and DS2 normalized from 0-100% of total stride time. Statistically significant differences data sets and gait symmetry represented by *.

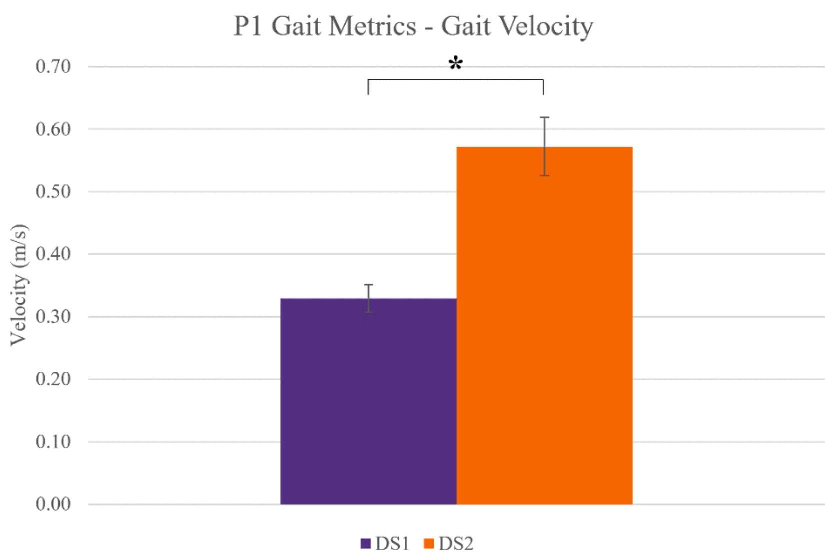

Figure 6. Gait velocity for DS1 and DS2. Statistically significant differences in data sets represented by *.

Patient 2 (P2)

The second patient was a Caucasian male, 56 years old, who suffered from a mild stroke. Two data sets were collected for P2, data set 1 (DS1) and data set 2 (DS2) on Day

1 and Day 7, respectively. Visually, there is little difference in pelvic tilt, obliquity, and rotation waveforms (Figure 7). Quantifiably, there was a decrease in the range of motion for all rotations of the pelvis, though none were significant. Both left and right hip flexion/extension movements produced a significant decrease in range of motion values between DS1 and DS2. There was no significant difference in range of motion between the left and right hip movements (Table 5).

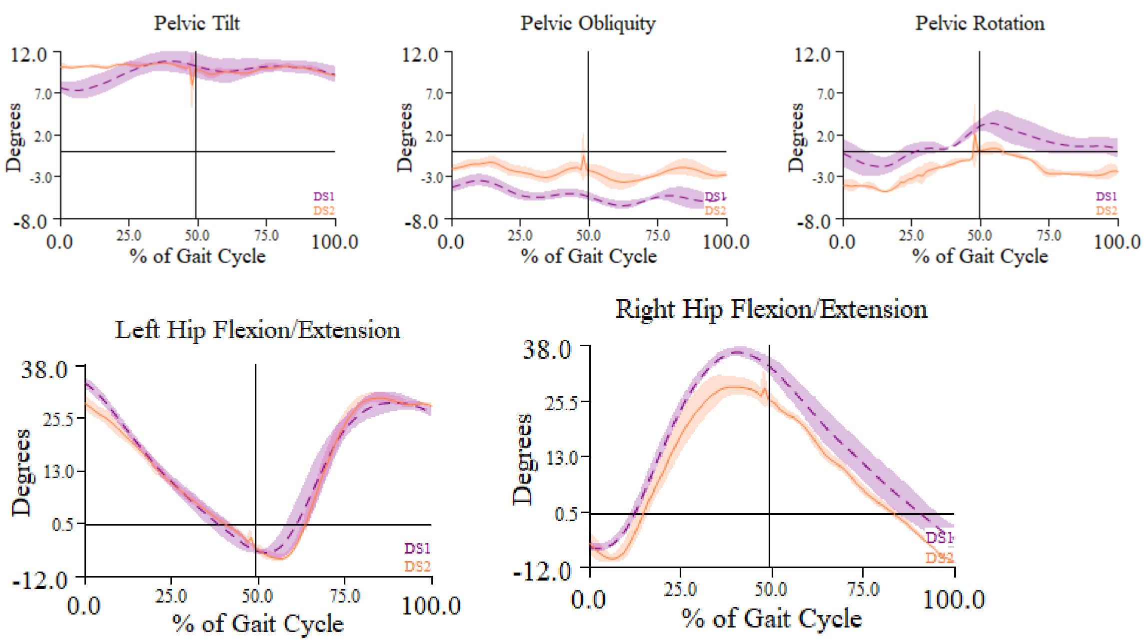

Figure 7. Kinematic waveforms of P2 pelvic tilt (top left), pelvic obliquity (top middle), pelvic rotation (top right), left hip flexion (+) and extension (-) (bottom left), and right hip flexion (+) and extension (-) (bottom right) during their gait stride, for DS1 (purple, dashed) and DS2 (orange, solid).

Table 5. Quantitative values for range of motion in the desired kinematic assessments. Statistically significant differences were present between DS1 and DS2 (*).

Data Set	Pelvic Tilt	Pelvic Obliquity	Pelvic Rotation	Left Hip Flexion/Extension	Right Hip Flexion/Extension
DS1	3.67±0.40*	3.22±1.07	5.51±0.13	40.73±1.84*	44.60±1.77*
DS2	1.70±0.10*	2.03±0.30	4.59±0.69	36.11±0.35*	38.97±2.66*

Gait metrics, stride length and stride time, were normalized 0-100% of the stride and divided between left and right steps. There was an increase in left step length between DS1 and DS2 and decrease in right step length, respectively, and both DS1 and DS2 showed significant differences in gait symmetry between left and right step lengths (Figure 8). However, this decreased step length symmetry from DS1 to DS2. It was also found that there was no significant difference among left step time or right step time between DS1 and DS2, although left step time increased while right step time slightly increased causing greater step length asymmetry in data sets. Similarly, no significant difference in step time was found between the left and right steps in either DS1 and DS2 (Figure 9). Overall gait velocity was found to significantly improve between DS1 and DS2 gait cycles (Figure 10).

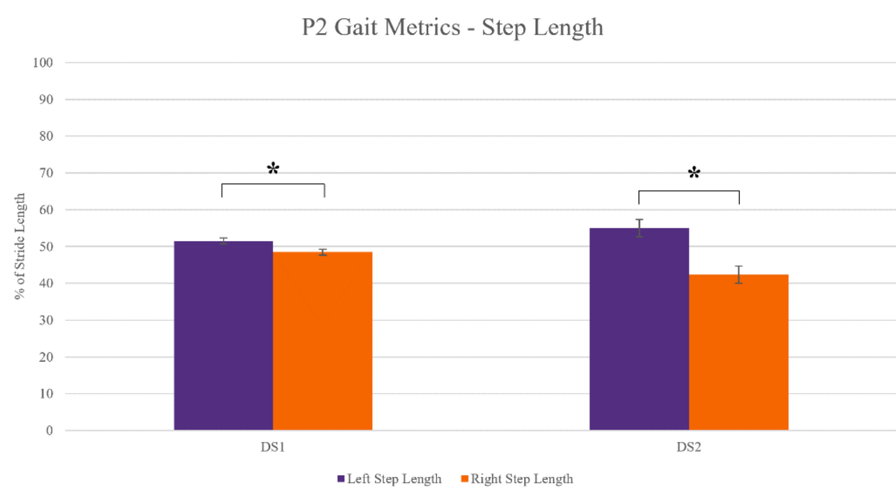

Figure 8. Gait step lengths for DS1 and DS2 normalized from 0-100% of total stride length. Statistically significant differences data sets and gait symmetry represented by *.

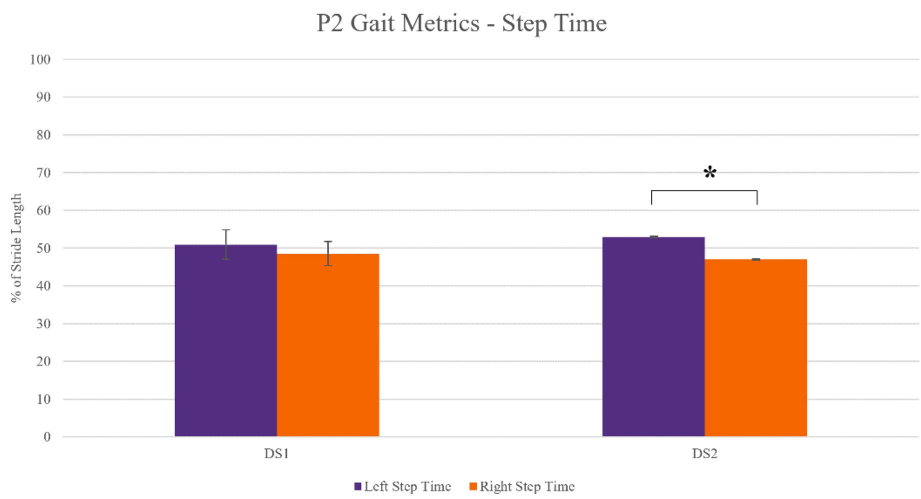

Figure 9. Gait step time for DS1 and DS2 normalized from 0-100% of total stride time. Statistically significant differences data sets and gait symmetry represented by *.

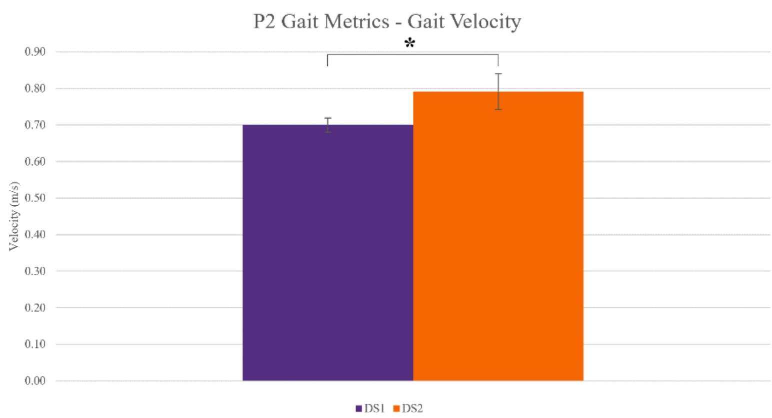

Figure 10. Gait velocity for DS1 and DS2. Statistically significant differences in data sets represented by *.

Patient 3 (P3)

The third patient was a Caucasian male, 54 years old, and suffered from a severe stroke. Four data sets were collected for P3, data set 1 (DS1), data set 2 (DS2), data set 3 (DS3), and data set 4 (DS4) on Days 1, 3, 7, and 9 respectively. Visually, results for pelvic tilt, obliquity, and rotation show consistent patterns, though no visible differences. Larger standard deviations were evident due to assistance from a walker (Figure 11). There was no significant changes in ranges of motion for the pelvic rotations. However, there was a significant increase in range of motion for the left hip and right hip flexion/extension patterns from DS1 to DS4. More, the left hip flexion/extension movement was significantly larger than the right in DS1 and DS4 (Table 6).

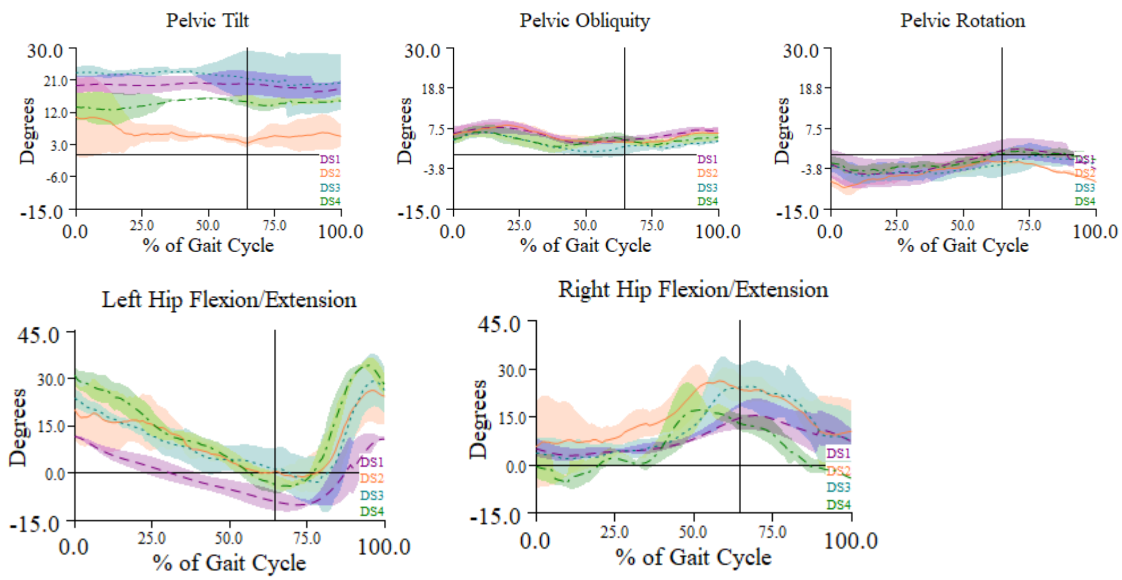

Figure 11. Kinematic waveforms of P2 pelvic tilt (top left), pelvic obliquity (top middle), pelvic rotation (top right), left hip flexion (+) and extension (-) (bottom left), and right hip flexion (+) and extension (-) (bottom right) during their gait stride, for DS1 (purple, dashed) DS2 (orange, solid), DS3 (blue, dot), and DS3 (green, dash-dot).

Table 6. Quantitative values for range of motion in the desired kinematic assessments. Statistically significant differences were present between DS1 and DS2 (*) and between right and left measurements (**).

Data Set	Pelvic Tilt	Pelvic Obliquity	Pelvic Rotation	Left Hip Flexion/Extension	Right Hip Flexion/Extension
DS1	2.46±0.82	5.17±1.97	6.85±0.82	23.72±2.27*,**	12.67±2.56*,**
DS4	3.94±4.70	3.77±1.01	5.20±0.66	37.44±4.38*,**	21.85±2.93*,**

Gait metrics, stride length and stride time, were normalized 0-100% of the stride and divided between left and right steps. Between DS1 and DS4, left step lengths significantly increased and, consequently, right step length significantly decreased. As a result, more symmetrical gait patterns were seen in DS4 than DS1. However, there was still a significant difference between right and left step lengths during DS1 and DS4

(Figure 12). Similarly, a significant difference in step time was found between the right and left step times for DS1 and DS4 data sets. Right step time was significantly larger than the left for both data sets. Right step time decreased while left step time increased, resulting in more symmetrical gait step times, however, no significant improvements in step time were found between DS1 and DS4 (Figure 13). Overall gait velocity was found to significantly improve between DS1 and DS4 gait cycles (Figure 14).

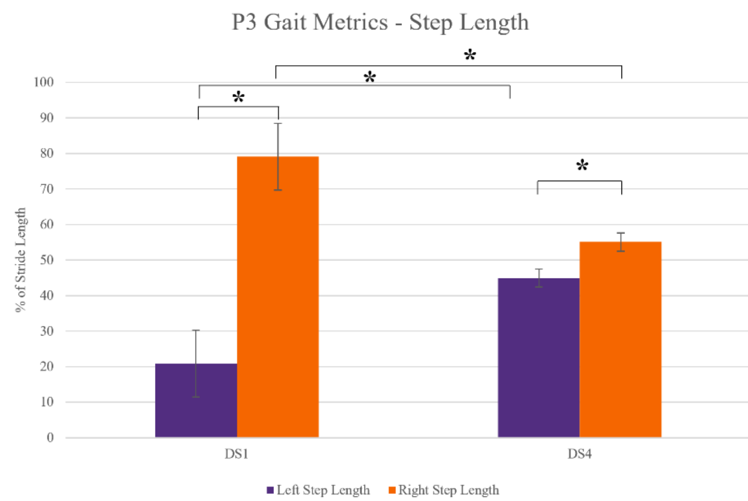

Figure 12. Gait step lengths for DS1 and DS4 normalized from 0-100% of total stride length. Statistically significant differences data sets and gait symmetry represented by *.

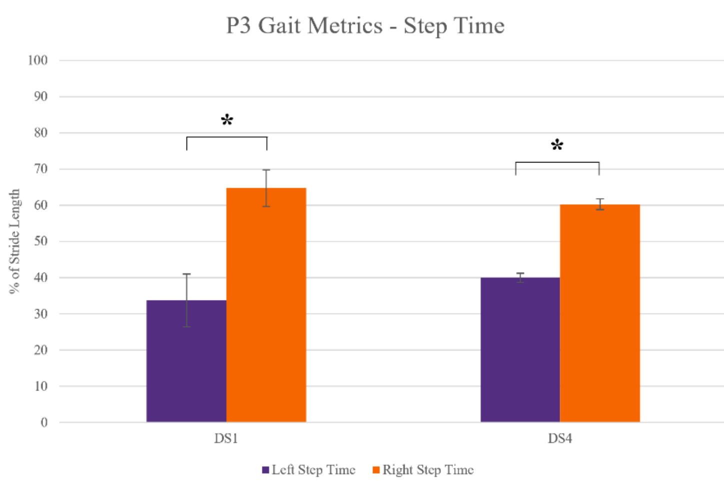

Figure 13. Gait step time for DS1 and DS4 normalized from 0-100% of total stride time. Statistically significant differences data sets and gait symmetry represented by *.

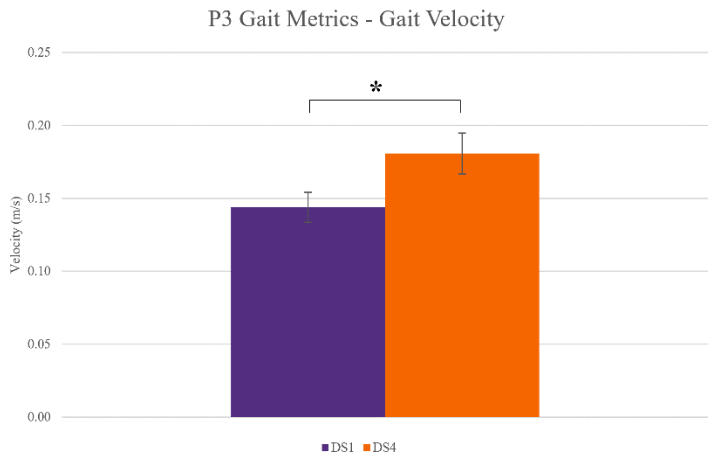

Figure 14. Gait velocity for DS1 and DS4. Statistically significant differences in data sets represented by *.

DISCUSSION

Pelvis and Hip Kinematic Discussion

Generally accepted terms for pelvic movements about the three cardinal axes are pelvic tilt, obliquity, and rotation. Pelvic tilt refers to the rotation about the mediolateral axis, or movement within the sagittal plane (Baker, 2001a; Lewis, Laudicina, Khuu, & Loverro, n.d.). Anterior pelvic tilt describes the movement of the anterior superior iliac spines (ASIS) targets in the anterior direction (Levangie & Norkin, 2011; M. Murray, Kory, & Sepic, 1970; Nuemann, 2017), creating a negative rotational movement for this model. Contrary, when the ASIS targets move posteriorly, the rotation about the axis is positive. Pelvic obliquity refers to the rotation about the anteroposterior axis, or movement within the frontal plane (Baker, 2001a; Lewis et al., n.d.). As the left side of the pelvis raises, or hikes, this produces a positive rotation within the model axis, and as the left side drops it creates a negative rotation. Rotation about the vertical axis, or within the transverse plane, is considered pelvic rotation (Baker, 2001a; Lewis et al., n.d.).

Backward or anterior movement of the left hip joint results in a positive rotation about the axis, and conversely for forward or posterior movement of the left hip joint results in negative rotation.

While dependent on axis orientation, the kinematic results for P1, P2, and P3 expressed similar waveforms for pelvic tilt, obliquity, rotation and hip flexion/extension as found in literature (Baker, 2001a; Bruening, Frimenko, Goodyear, Bowden, & Fullenkamp, 2015; Crosbie, Vachalathiti, & Smith, 1997; Kadaba, Ramakrishnan, & Wootten, 1990; Lewis et al., n.d.; M. Murray et al., 1970; Smith, Lelas, & Kerrigan, 2004). More visible in P1 and P2 pelvic tilt waveforms, is the initial posterior (positive) tilt of the pelvis, followed by an anterior (negative) tilt at approximately 50% of the gait cycle, and then posterior and anterior rotations again during the remainder of the gait cycle (Bruening et al., 2015; Crosbie et al., 1997; Kadaba et al., 1990; M. P. Murray, Drought, & Kory, 1964; Smith et al., 2004). While evident, this movement is slightly delayed in P1 and P2 waveforms, likely due to impaired gait and delayed response of the pelvis following foot strikes and stabilization. More, P3 does not display the exact described pattern, however, this patient was considered more severely impaired and required assistance while walking and consequently did not create representative waveforms. More, all three patients pelvic tilt total mean ranges of motion were consistent to those found in literature (Table 4, Table 5, Table 6) (Bruening et al., 2015; Crosbie et al., 1997; Lewis et al., n.d.; M. P. Murray et al., 1964; Smith et al., 2004). Similarly, pelvic obliquity patterns were found to be consistent with those in literature, performing two cycles of the pelvis rotating positively, or hiking, at initial step contacts and rotating negatively, dropping, during the contralateral leg swing (Baker, 2001a; Levangie & Norkin, 2011). Desired obliquities mean range is 6-11 degrees on healthy human gait, and P1's values fell within this range (Bruening et al., 2015; Crosbie et al., 1997; Lewis et al., n.d.; M. P. Murray et al., 1964; Smith et al., 2004). While P2's and P3's mean range was slightly smaller in value, this is likely due to their impaired gait that results in limited joint range of motion.

Positive rotation of the pelvis occurs briefly after the left heel strike, as the right leg begins its swing phase. This positive rotation continues until the contralateral leg, right, creates contact and the left leg begins its swing phase resulting in a negative pelvic rotation (Baker, 2001a; Lewis et al., n.d.). The mean range for pelvic rotation is 3-14 degrees, and all patients fell within the lower range of these magnitudes likely due to their slower gait (Bruening et al., 2015; Crosbie et al., 1997; Lewis et al., n.d.; M. P. Murray et al., 1964; Smith et al., 2004). Hip flexion and extension was also consistent with gait patterns and mean ranges for impaired gait (Carmo, Kleiner, Lobo da Costa, & Barros, 2012; Chen, Patten, Kothari, & Zajac, n.d.).

Gait Metrics Discussion

Individual results for P1, P2, and P3 show improvements in asymmetrical gait and velocity. P1 exhibited a statistically significant smaller step length on the right side than the left, while also exhibiting a significantly larger hip flexion/extension range on the right hip than the left. Step length is considered to be the distance between the proximal end position of the contralateral foot at the previous heel strike to the proximal end of the ipsilateral foot at the next heel strike. Therefore, the ipsilateral step length could be significantly smaller than the contrary, but the patient wouldn't need a large extension of their contralateral hip in order to maintain their stride. P1's data set expresses this pattern for DS1, however, by DS2 there was no significant difference in hip flexion/extension between sides. There was still a significant difference in step length, however, asymmetry between these values greatly improved. Similarly, P3 exhibited significantly larger step lengths on the right side initially, which resulted in significantly larger hip flexion/extension range on the left side of their stride. Asymmetry in step length improved by DS4, however, differences in step length and hip flexion/extension were still significantly relevant.

There is a strong correlation found between hip flexor strength and improvements in gait velocity (Hsu, Tang, & Jan, 2003; Maria Kim & Eng, 2003; Nadeau, Arsenault, Gravel, & Bourbonnais, 1999). In this study, mean hip peak flexion/extension was tracked for each patient and statistically significant increases for this range were found in P1 and P3, while P2 had a significant decrease in range. Both P1 and P3 also had statistically significant improvements in gait velocity between their first and final data sets, as did P2. While more data is needed to perform a statistically powerful model, it can be inferred that increased hip flexor strength would consequently improve the range of motion in hip flexion/extension. Therefore, there may also be a connection between improvements in hip flexion/extension and gait velocity, as was seen in P1 and P3 results.

From the clinical outcome perspective, there were significant improvements in step length for P1 and P3 from their initial data sets to their final assessment. Even more for these two patients, their gait symmetry also improved. These results, coupled with their significant increase in gait velocity, indicate expansive growth in gait metrics during their inpatient stay at the hospital, and gains during their spontaneous recovery phase. P2 did not generate significant improvements in step length or step time, but did in gait velocity. Initially, this patient expressed the least severity in gait abnormality due to their stroke impairment and it's possible that additional data sets would have provided more intrinsic data to this anomaly. Additional data sets on more patients will be collected to strengthen the overall data findings. However, with the current findings, the assessment found positive improvements in the gait-focused rehabilitation approach at the current hospital.

Common rehabilitation assessments post-stroke include the Canadian Neurological Scale (CNS) (Côté et al., 1989), Barthel Index (Jamemucu, 2003), Balance Scale (Berg, 2009; Berg, Wood-Dauphinee, & Williams, 1995), TUG (Podsiadlo & Richardson, 1991), Box and Block Test (Desrosiers, Bravo, Hébert, Dutil, & Mercier, 1994), Stroke Rehabilitation Assessment of Movement (STREAM) (Daley, Mayo, & Wood-Dauphinée, 1999) and gait speed measurement (Salbach et al., 2001). The focus on motor recovery is a common theme in any stroke assessment and its critical to assess patients properly in order to predict patient discharge and, hopefully, minimize their length of stay in the hospital (Daley et al., 1999). These standard practices assess similar measurements found in this study. However, these assessments require a form of measurement that is subjective to the administering personnel and are primarily focused on one functional motor ability, such as gait speed. While most stroke patients who suffer from mobility impairment are expected to see improvements (Bonita & Beaglehole, 1988), the assessment results can vary dependent on who is administering the test and what test was chosen.

CONCLUSION

The improvements seen in this study show results for a successful patient rehabilitation approach. More, the use of video motion capture technologies show objective data collection methods that encompass a larger data set than current assessments, to better determine patient recovery during their inpatient stay. These results are generated more efficiently that current approaches that require multiple, different assessments. As motion capture and EMG technologies become more financially available, clinical settings could permanently implement these methods as a standard practice.

CHAPTER TWO

INFLUENCE OF 8-WEEK HORSEBACK RIDING ACTIVITY ON BALANCE AND PELVIC MOVEMENTS IN AN OLDER ADULT POPULATION

INTRODUCTION

Balance and gait are key indicators of the overall health of older adult individuals, as these are strongly linked to vision and the nervous and musculoskeletal systems (Horak, 1997). A cross-sectional analysis of the 2008 National Health Interview Survey found that approximately 20% of older adult citizens in the United States experience problems with dizziness or balance annually (Lin & Bhattacharyya, 2012). A 2019 literature review of articles (1980-2020) pertaining to gait and balance concluded that older adults experience an age-dependent decline in balance, and this in turn causes gait abnormalities and an increased fall risk (Osoba, Rao, Agrawal, & Lalwani, 2019). This is a significant health issue, as 35% of individuals aged 70 and older have gait disorders, and falls are the most common cause of accidental death or injury in this population (Center for Disease Control and Prevention, 2016).

This high prevalence of balance related issues has made balance coordination-related exercises a key priority in modern physical rehabilitation (Araujo, Silva, Costa, Pereira, & Safons, 2011). Frequent and systematic exercise has been found to be effective in combating these negative effects of ageing (Christmas & Andersen, 2000a; Nicola & Catherine, 2011; Peterson, Rhea, Sen, & Gordon, 2010; S & T, 2006). Hippotherapy, or forms of therapeutic horseback riding and activities, is an alternative to traditional exercise that is used in some rehabilitation programs to improve balance and gait (American Hippotherapy Associations, n.d.; Araujo et al., 2011; Bronson, Brewerton, Ong, Palanca, & Sullivan, 2010; Sterba, 2007). The three-dimensional motions of a walking horse provoke full body, cyclic movement in a rider that can help develop strength, balance, and coordination (Benda, Mcgibbon, & Grant, 2003; Janura, Peham, Dvorakova, & Elfmark, 2009a). Hippotherapy can be particularly helpful in older adult populations, as it has been shown that specific structured exercise programs are more likely to motivate participation of independent older adults to improve balance and strength (Silveira, Reve, Daniel, Casati, & De Bruin, 2013). Furthermore, horseback riding allows riders to experience 3D motion patterns that sometimes cannot be produced independently, which can lead to increased functional improvements (Beinotti, Correia, Christofoletti, & Borges, 2010). More specifically, a 2014 study showed that riding elicited pelvic motion patterns similar to standard gait in the transverse and frontal planes using 3D motion capture in young, healthy subjects (Garner & Rigby, 2015a).

Multiple studies have been conducted to evaluate the effects of hippotherapy on gait and balance improvements. A 2014 study found that a group of healthy older adult subjects had a significant decrease in gait sway and improvement in static balance in comparison to a treadmill control group using motion capture and a balance performance monitoring force plate (M. Lee, Kim, & Park, 2014). Another group found that eight weeks of horseback riding led to improved lower limb strength and balance using a 30 second chair stand test and the Berg Balance Scale (Araujo et al., 2013). In an earlier study, this same group found that after sixteen horseback riding sessions, there was a significant improvement in timed up and go which is a predictor of fall risk (Araujo et al., 2011). In addition to the methods used to quantify balance in these studies, the Fullerton

Advanced Balance (FAB) scale is widely used to measure improvements in balance (Gouveia et al., 2014; Hernandez & Rose, 2008a; Hoskovcová et al., n.d.). This scale was found to be highly reliable upon its conception in 2006 when validated against previous literature for balance assessments (Rose, Lucchese, & Wiersma, 2006).

While current evidence supports the positive influence of hippotherapy on balance, the biomechanics of the rider and horse during horseback riding that produces these improvements is largely unexplored. While all used motion capture technologies to track rider kinematics (Beinotti et al., 2010; Byström, Rhodin, Peinen, Weishaupt, & Roepstorff, 2009; Garner & Rigby, 2015a; Münz, Eckardt, & Witte, 2014a), they did not investigate the relationship between horse and rider movement, and discuss these kinematic interventions with respect to a larger health related outcome such as balance. By defining the kinematic relationship between the horse and rider, methodologies and models could be developed and implemented to specific patient populations. Therefore, the purpose of this study was to quantify the basic kinematic relationships between horse and rider movements in a healthy cohort. It was hypothesized that during a hippotherapy-like walking session, the four-beat gait pattern of the horse's stride will result in a double-pitching pattern of the rider's pelvis that is kinematically relevant to balance training interventions. Secondarily, this study also aimed to relate this riding intervention to measured changes in balance following an eight-week horseback riding regimen in an older adult cohort. The stimulation of the flexion and extension of the rider's hips may have implications in their balance and gait.

RESEARCH METHODS

Animal Care

All studies were conducted with IACUC approved protocols. Sound, gentle horses with previous EAAT experience were used for every riding session with unexperienced riders. The horse's health was consistently be inspected by the center's staff. Further, riders were instructed and assisted by team members while mounting, riding, and dismounting in order to approach each task with the horse's safety as a priority. All horseback riding sessions took place at the Clemson University Equine Center.

Study Design and Patient Recruitment

Following institutional IRB approval, ten participants (range 60-80 yrs, mean 67.7±3.5, 8F/2M) with no-to-moderate riding experience were chosen from a community solicitation. Following institutional IACUC approval, two horses of different conformation types were chosen. One a female (15.3 hands, 20 years old) and one male (16 hands, 19 years old) horses were used with consistent riders throughout the eight weeks. Both horses were gentle natured and had prior experience in a therapeutic riding setting. Participants signed an informed consent and lessons were held at the University Equine center in their outdoor arena (100x200 ft). A 30 ft walkway was created with GoPro (San Mateo, CA) cameras (300 Hz) set up 30 ft across from each other (Figure 1).

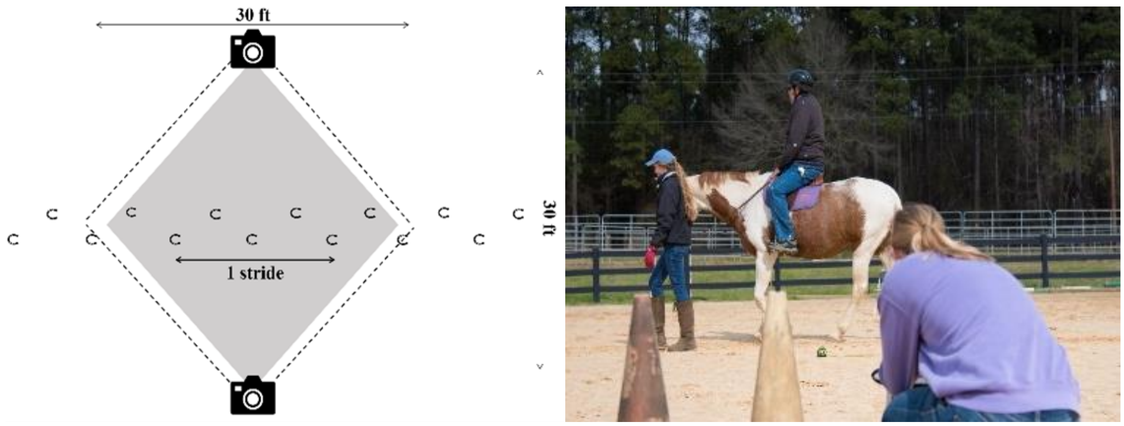

Figure 15. Data collection motion capture volume (left) and representation of horse's stride length during capture. Participant led at a walk through the motion capture volume (right).

FAB assessments were conducted by an experienced recreational therapy team at the university who had conducted FAB assessments in the past. Participants took the FAB balances assessment on separate days prior to their first ride and following their last ride. During the first and last rides, a video motion capture system (GoPro, California, USA) was set up around the walkway to record the horse and rider, during 5 separate passes. Each horse was suited with markers at the distal femur head, greater trochanter, and point of hip (Figure 16). These three markers create the hip joint connecting the horse's pelvis to their femur. The rider's movements were tracked with single and dual axis goniometers (Biometrics, LTD, United Kingdom), with a tolerance of +/- 2°. The rider sensors were placed across the right and left hip, and thoracic and lumbar spine (Figure 16). Outside of data collection, participants received 30-minute riding lesson once a week for eight weeks, with the horse led at a walk. Lesson instructors developed riding lesson patterns and skills based on individual growth and comfort levels.

Data Processing

Markers on the horses were tracked (300 Hz) for one stride and analyzed within MATLAB (Hedrick, 2008). Rider joint angles at the left and right hip and femur were tracked for passes 1-5 for each rider during week 1 and week 8 (Figure 16). This data was exported into Microsoft Excel where marker locations were converted into vectors based on their two-dimensional location in each individual frame, and the joining of the two vectors created the anatomical angle, α_R. The continuous angle data was quantified by the Law of Cosines (Equation 1) and was used to obtain range of motion values at the horse's hip throughout the duration of their stride. Strides were exported from Excel into a custom MATLAB program to filter, normalize 0-100% of the gait cycle, and export data back into Excel. For each horse (A and B), each α range during the entire stride cycle, left and right, was averaged across all passes for all corresponding riders. The average pelvis femur range of motion pattern, α_H, was graphed with standard deviations for week 1 and week 8 as representative for 2D.

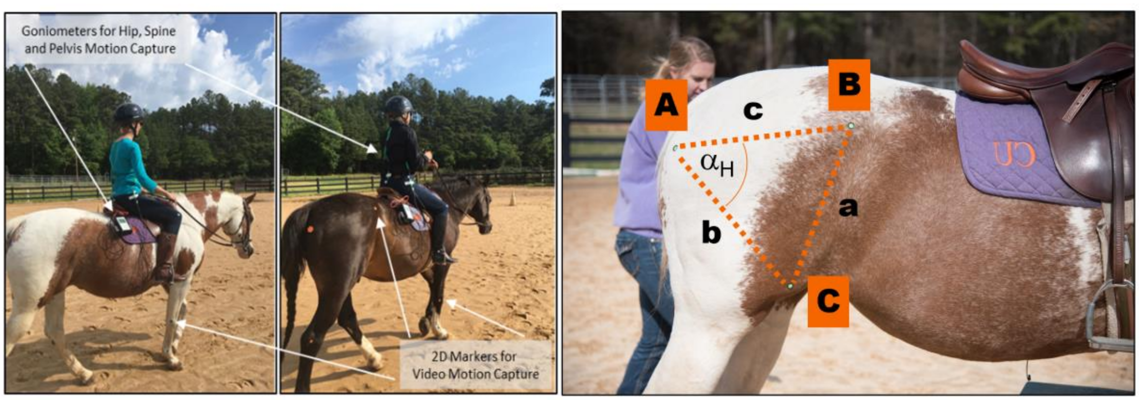

Figure 16. Dual axis goniometer system attached to rider's hips and target markers for hip flexion/extension analysis of the horse (left). Desired pelvis-femur angle (α_H) for analysis based on target markers at anatomical locations on the horse (right).

Table 7. Kinematic parameters tracked for horse and rider analysis during the horse's gait, α_R and α_H. Balance measurements were quantified using the Fullerton Advanced Balance assessment

Parameter	Method	Method Units
Horse hip flexion/extension (α_H)	MATLAB GUI	Position (X,Y)
Rider hip flexion/extension (α_R)	Biometrics dual-axis goniometer	Degrees (°)
Rider balance assessment	Fullerton Advanced Balance Scale	Assessment scoring 0-4/per exercise

$$a^2 = b^2 + c^2 - 2bc\cos\alpha$$

Equation 1. Law of Cosines

$$a = \sqrt{(Y_{Ci} - Y_{Bi})^2 + (Z_{Cj} - Z_{Bj})^2}$$

Equation 2. Side (a) distance from point of hip to distal femur head

$$b = \sqrt{(Y_{Ai} - Y_{Ci})^2 + (Z_{Aj} - Z_{Cj})^2}$$

Equation 3. Side (b) distance between point of hip to greater trochanter

$$c = \sqrt{(Y_{Ai} - Y_{Bi})^2 + (Z_{Aj} - Z_{Bj})^2}$$

Equation 4. Side (c) distance between greater trochanter and the distal femur head

The riders were separated by the horse they rode during the entirety of the study, riders 1A-5A and riders 1B-5B. Hip flexion and extension goniometer data was pulled and averaged across the entire horse stride for 5 passes for each rider during week 1 and week 8 separately. Similarly, goniometer data was exported from the Biometrics software into Excel, divided into strides, and normalized 0-100% of the gait cycle, and filtered within the custom MATLAB script (MathWorks V.17a). Individual riders' range of motion during their flexion and extension pattern of an equine stride was averaged across five passes. Their average range of motion was analyzed in a two-tailed paired t-test (α=0.05) to compare left and right range imbalances and week 1 and week 8 variations. Riders' average flexion and extension pattern were averaged for week 1 and week 8, between horse A and B. These averaged patterns were graphed across the gait cycle and considered representative of the 2D cyclic pelvis pattern experienced by the rider.

RESULTS

Kinematic Results

Repeatable and cyclic rider movement patterns were captured with goniometers. Two pitching cycles of the riders' pelvis per horse walking cycle were evident in all riders. Summed results for Horse A are seen in Figure 17. Summed results of all Horse B riders are shown in Figure 18. Individual results of the riders' average hip flexion/extension range of motion (left and right) with their standard deviations across five passes were calculated but no significant pattern was found (Appendix A).

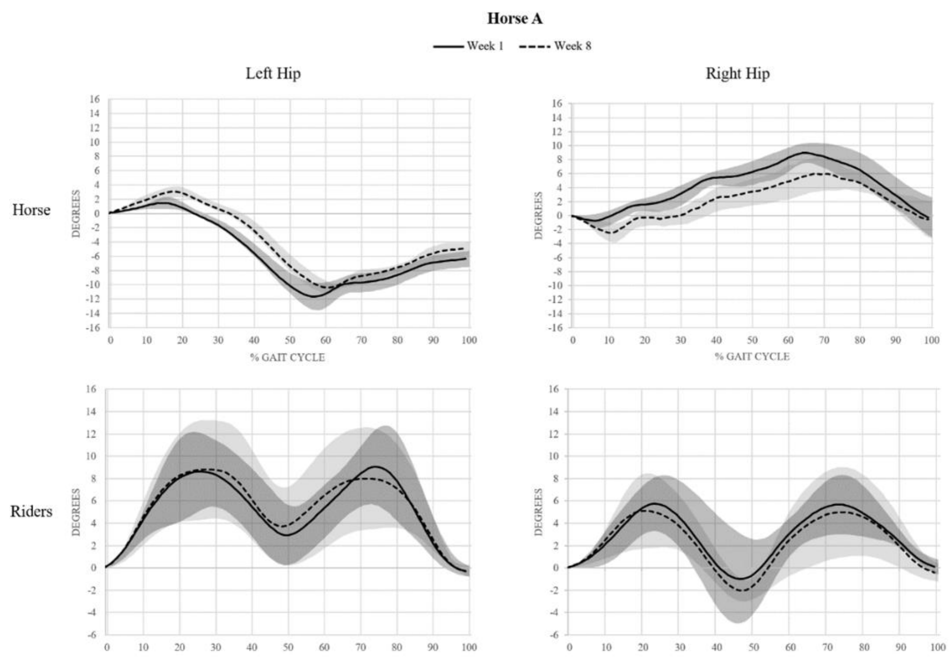

Figure 17. Horse A average hip flexion and extension patterns during their gait stride (top) between week 1 (solid) and week 8 (dashed). Riders' averaged hip flexion and extension patterns during the horse's gait stride (bottom) between week 1 (solid) and week 8 (dashed).

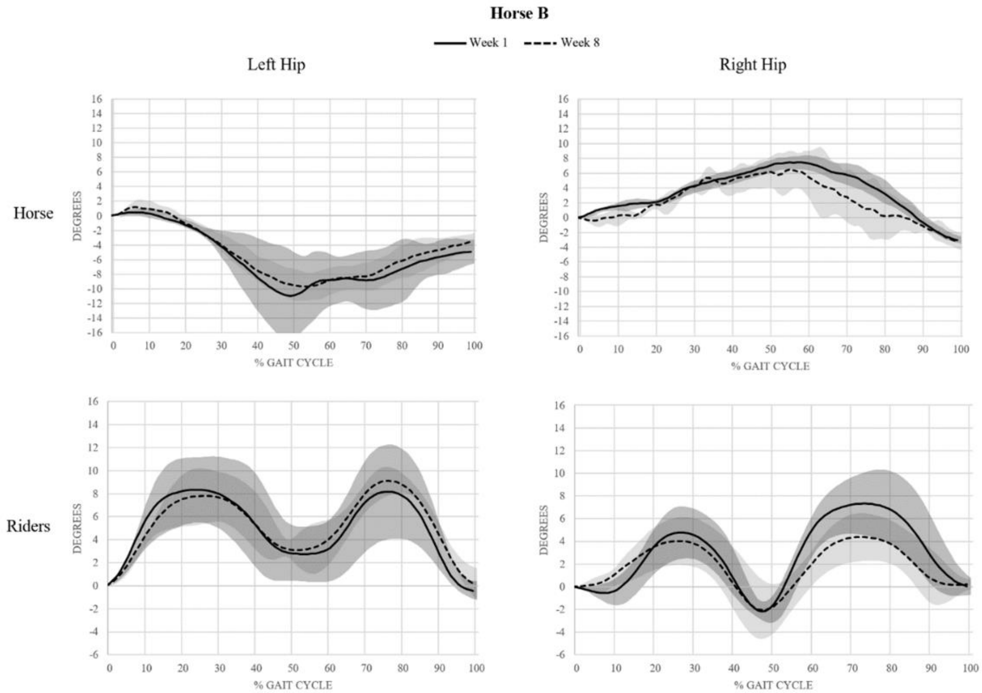

Figure 18. Horse B average hip flexion and extension patterns during their gait stride (top) between week 1 (solid) and week 8 (dashed). Riders' averaged hip flexion and extension patterns during the horse's gait stride (bottom) between week 1 (solid) and week 8 (dashed).

In addition, horse pelvis-femur joint angles were tracked across a stride and averaged across all riders, for both horse A and B. The results of these stride patterns are seen in Figures 4 and 5. Both horse A and B show similar stride movement patterns and ranges of motion in the specific joint, expected at a walk.

In general, equine left hip pitch (or extension) occurred between 6-18% in the gait cycle, and the right hip pitch peaks occurred between 55-68% in the gait cycle. This resulted in a total range of motion in Horse A of 13.12° and 9. 68°, and 13.42° and 8.42° for the left hip and right hip (weeks 1 and 8) respectively, in addition to 11.44° and 10.48°, and 10.91° and 9.74° for equivalent measurements in Horse B. This produced rider pelvic pitching at 24-28% and 72-75% of the gait cycle. Slight differences in horse hip kinematic range of motion and peak pitch phasing were seen between horses and weeks but none were statistically significant. Rider peak pelvic pitch response characteristically lagged the peak hip pitch of the horse by 8-20% as the kinematic impulse was transmitted from the horse to the rider. Both the left and right hip of the rider pitched in response to either the left or right horse hip peak motion, which produced a characteristic two beat sway of the rider on the horse for each horse gait cycle. During these riding lessons, this rider hip movement occurred approximately every 1-1.5 seconds.

These movement patterns showed one peak and one valley on both the left and right sides during one stride of both horses. This double pitching undulation effect of the combined left and right pelvis-femur was evident in the double peaks in the riders' pelvic flexion and extension, and the rider's peaks occur delayed to the horses'. A representative movement pattern from each horse is shown in Figure 17and Figure 18.

Total range of motion of five-pass averages of the pelvis rotating in the pitching direction were calculated and compared between week 1 and 8 and right and left sides for all riders. Paired two sample t-tests were used to analyze differences between right and left hips, across week 1 and 8. While some riders significantly increased or decreased their total average range of motion in their riding pattern, there was no consistent pattern across all riders. It was expected that riders with very little horseback riding experience would exhibit larger movement patterns initially and then a decrease in their pelvis mobility (range of motion) as they grew more comfortable horseback by week 8. However, this was not seen within the data sets. Some novice riders exhibited less movement initially, possibly due to the cold weather and rider hesitancy in week 1, and then showed increases in their riding pattern motion range by week 8. A few of the more experienced riders showed decreases in rider movement between weeks 1 and 8, possibly due to increased familiarity with the horses movement. Intra-variability of each rider's multiple passes was small with the greatest standard deviation being 4.79° in the right hip for Rider 3B during week 8. These values are shown tabulated (Table 8).

Table 8. Range of motion for riders 1-4 (R1A-R4A) pelvis pitch patterns of the right hip (RH) and left hip (LH), for a single stride in Horse A. Range of motion week 1 (W1) and week 8 (W8) and their corresponding standard deviations (St. Dev.) are in degrees. Range of motion for riders 1-3, and 5 (R1B-3B, R5B) pelvis pitch patterns of the right hip (RH) and left hip (LH), for a single stride in Horse B. Range of motion week 1 (W1) and week 8 (W8) and their corresponding standard deviations (St. Dev.) are in degrees.

Riders (A) Range of Motion

R1A	LH	RH	P(T<=t)
W1	8.57±1.55	9.07±0.72	0.475
W8	8.76±1.48	9.53±1.25	0.325
P(T<=t)	0.88	0.43	
R2A	**LH**	**RH**	**P(T<=t)**
W1	11.51±2.07	7.58±0.94	0.007
W8	9.01±1.76	5.44±0.61	0.036
P(T<=t)	0.124	0.036	
R3A	**LH**	**RH**	**P(T<=t)**
W1	16.59±0.89	9.28±1.00	0
W8	6.18±0.91	6.11±0.29	0.884
P(T<=t)	0	0.004	
R4A	**LH**	**RH**	**P(T<=t)**
W1	8.99±1.78	9.74±1.17	0.5
W8	17.48±2.73	12.99±2.36	0.105
P(T<=t)	0.013	0.039	
SUM	**LH**	**RH**	
W1	11.42±3.19	8.92±0.81	
W8	10.36±4.26	8.52±3.01	

Riders (B) Range of Motion

R1B	LH	RH	P(T<=t)
W1	8.16±1.68	8.17±0.59	0.992
W8	10.2±1.71	6.73±1.4	0.02
P(T<=t)	0.126	0.109	
R2B	**LH**	**RH**	**P(T<=t)**
W1	9.75±1.43	12.01±2.13	0.113
W8	10.71±0.9	7.69±1.37	0.005
P(T<=t)	0.375	0.001	
R3B	**LH**	**RH**	**P(T<=t)**
W1	9.28±1.3	13.87±1.33	0.001
W8	12.02±1.09	12.77±4.79	0.742
P(T<=t)	0.054	0.656	
R5B	**LH**	**RH**	**P(T<=t)**
W1	10.66±0.71	7.34±1.16	0.018
W8	9.73±2.14	7.59±0.94	0.2
P(T<=t)	0.359	0.815	
Sum	**LH**	**RH**	
W1	9.46±0.90	10.35±2.69	
W8	10.67±0.86	8.70±2.34	

Balance Results

Due to the small-scale rating of the assessments (1-4), only exercises that had the chance to be improved upon (<4) were compared between week 1 and 8 assessments.

These results for week 1 and 8 were summed for all riders across exercises and summed for all exercises across riders. Paired two sample t-test for means was used to analyze the grouping differences. For both assessments, week 8 had significantly better scores than week 1. Scores improved with all riders from 73 to 96. Further, the sum of all the averages of the riders' scores were analyzed between week 1 and week 8 and significantly improved from 19.1+/-0.528 to 25.4+/-0.629. The sum of each exercise averages across all riders also significantly improved from 22.7+/-0.85 to 30.8+/-0.84.

Interestingly, participant 2A had an original FAB score of 24, mean of 2.4+/-1.4, which was considered under the threshold (25) as a predictor of increased fall risk in older subjects (Hernandez & Rose, 2008b). This rider improved to a total score of 34, mean score 3.4+/-0.8, resulting in a significant FAB mean score increase ($p(T<=t)=0.0160$ with $\alpha=0.05$). All other riders did not have initial FAB scores that fell below the summed score of 25 and were assessed as previously described.

Table 9. Fullerton Advanced Balance (FAB) assessment statistical analysis results and statistical significance was found when comparing Rider Sums, Rider Averages, Exercise Sums, and Exercise Averages between week 1 and week 8 (grey).

Fullerton Advanced Balance Assessment Scores

FAB (scores <4 in week 1)	W1	W8	P(T<=t) α=0.05
Rider Sums	73.0	96.0	0.018
Rider Averages	19.1+/-0.52	25.4+/-0.62	0.001
Exercise Sums	73.0	96.0	0.024
Exercise Averages	22.7+/-0.85	30.8+/-0.84	0.006

DISCUSSION

Kinematic Discussion

Cyclic motion patterns of the rider's pelvis pitching motion and the horse's pelvis-femur flexion and extension were successfully captured, and demonstrated that the four-beat gait pattern of the horse's stride resulted in a double-pitching pattern of the rider's pelvis. This pattern was highly repeatable with most riders, and measures of ranges of motion within individual riders was low. In fact, with the reported goniometer repeatability being 1°, and the accuracy being +/-2°, only six of the thirty-six reported measures of average ranges of motion showed greater than a +/-2° standard deviation (Table 8).

Rider pelvic range of motion is of specific interest in this study. Two studies have found that a rider's pelvis rotates posteriorly between contact of the horse's hind and forelimb and anteriorly between fore and hind limb contact (Byström, Rhodin, von Peinen, Weishaupt, & Roepstorff, 2009a; Münz, Eckardt, & Witte, 2014b). The stride patterns of the horses' left joint agreed well to the frontal plane movement patterns of the simulator and rider response found in a study (Goodworth, Barrett, Rylander, & Garner, 2018a). However, the horses' right joint clearly reflected an inverse movement pattern because of the defined stride. If the pelvis-femur joint directly induced the frontal pelvis motion, then it likely represented the right and left kinematic patterns. However, the left side differed from the frontal movement. The simulator (Goodworth et al., 2018a) was similar to the stride kinematics of the horse in the current study. Although, the horse, as a skeletal structure, could act as a linkage system and the pelvis-femur joint could have an indirect effect on the frontal plan reaction of the rider. Additionally, "List Angle" tracked by (Garner & Rigby, 2015a), equivalent to the frontal plane oscillation, showed a general double peak with both occurring at approximately 17% and 83%, respectively. A representation of the right and left pelvis-femur joints, averaged between week 1 and 8 (Figures 4 and 5), showed a similar double peak movement pattern. These left pelvis peaks occurred at 15% (Horse A) and 8% (Horse B) and 65% (Horse A) and 57% (Horse B) for the right hip, during week 1. Horses have similar four-beat gait patterns at a walk and, thus the horses used in the (Garner & Rigby, 2015a) study by Garner likely had similar double peak flexion and extension pelvis-femur patterns in a single stride. The peaks in our horses' strides were slightly ahead of the list angle peaks in the referred study, however, this double undulation could support the belief that the pelvis-femur joint was linked to the point of movement translation to the frontal pelvis motion or list angle. This relationship can be explored in further studies.

Two pitching peaks of the pelvis were evident in all riders during one stride of the horse at a walk. This pelvic movement pattern was similar to others found in literature using both inertial measurement units (Eckardt & Witte, 2017a; Münz, Eckardt, Heipertz-Hengst, Peham, & Witte, 2013a; Wang et al., 2018a), and video motion capture (Byström, Rhodin, von Peinen, et al., 2009a; Goodworth et al., 2018a). The average flexion and extension of the riders' pelvis across horse A and B were similar to those values found in literature. The study by Wang (Wang et al., 2018a) found their participants exhibited 8.01±1.02 of pelvic flexion/extension for professional riders and 9.94±0.080 for beginners, while (Münz et al., 2013a) similarly found 11.1±3.6 for professionals and 8.1±4.1 for beginners. In addition, another study also found pelvis pitch range of motion to be 9.7±2.0 for the riders in their study. Across week 1 and 8, and both horses, the riders in this study showed very similar ranges of motion (Table 8).

There was no significant pattern for increase or decrease in range of motion values among the participants between week 1 and week 8 of riding, however the capabilities of tracking rider pelvis pitching pattern in relation to horseback riding is important to discuss. It has been shown that horseback riding has an effect on strength and coordination (Benda et al., 2003; Janura, Peham, Dvorakova, & Elfmark, 2009b). Therefore it can be hypothesized that that the two-peak pitching undulation pattern in this study's participants also had a similar impact on the riders..

The ranges of motion experienced by riders were similar to kinematically relevant movements in a clinical setting. A study by Chiacchiero (Chiacchiero, Dresely, Silva, DeLosReyes, & Vorik, 2010) assessed an equivalent motion in hip flexion and extension of older adult fallers and nonfallers using a goniometer based technique. Similarly, other studies have addressed the hip pitching (flexion/extension) motion in a clinical setting in older adults (Bello, Ababio, Antwi-Baffoe, Seidu, & Adjei, 2014; Kerrigan, Lee, Collins, Riley, & Lipsitz, 2001; Watt et al., 2011) as clinically relevant information for gait and balance health. These assessments were especially relevant when addressing balance disparities and age because of the direct relationship between hip flexion and extension and human gait. Our results were a representation of these clinically relevant hip patterns while horseback riding, suggesting that horseback riding can successfully achieve appropriate movement patterns simulated in the clinical setting for hip flexion and extension.

It was found that non-fallers had larger hip flexion and extension ranges of motion than fallers due to the decrease in stride length and/or the taut hip flexors that restrict full hip movement in fallers (Chiacchiero et al., 2010), and similarly for young adults than older adult non-fallers (Kerrigan et al., 2001). In addition, hip flexion and extension increased with gait speed in older adult fallers and non-fallers (Kerrigan et al., 2001). With a therapeutic horseback riding session, the rider might have the ability to control the overall speed of the horse, and this determines the rate of rhythmic movement based on the stride length of the horse. Therefore, the rider could use the horse as a platform to control the rate of gait-like movements that might be otherwise unattainable during normal rehabilitation. Therefore, we hypothesize that the desire for increased hip flexion and extension in the older adult population, to further promote balance and gait improvements, could be achieved during horseback riding if it cannot be accomplished on one's own.

Balance Discussion

The FAB has been found to be a a reliable and valid test for balance in community dwelling, older adults (Rose et al., 2006). As stated, rider 2A had significant improvements across eight weeks and the entirety of riders were analyzed based on their opportunity for improvements as a whole group. It was not unusual that the majority of riders did not show significant improvements across eight weeks, as (D. N. Homnick, Henning, Swain, & Homnick, n.d.) did not see significant improvements in mean balance scores until weeks 8-16 in a 24 week study. In addition, riders in this study only received

30 minutes of horseback riding per week. Increased horseback riding lessons or increased study length could have resulted in greater FAB improvements.

Additionally, the study population of 60-80 year old were on the healthier spectrum and, therefore, had higher initial FAB score assessments. Because of the limitations in the study, only measures of exercise performance that had opportunity for improvements between week 1 and 8 (those that were initial measured as less than 4 on a 4 point scale) were assessed for improvement analysis. The four assessments (rider sums, rider summed averages, exercise sums, and exercise summed averages) were analyzed and found to have significant improvements between week 1 and 8. Similar studies used functional balance measures to track the effects of hippotherapy and horseback riding simulators on balance in older adults. (De Araújo et al., 2013) participants were enrolled in 16 sessions of hippotherapy over eight weeks and found improved lower limb muscle strength and balance. Another study utilized a horseback riding simulator for a six week intervention on older adult participants with dementia and found significant improvements in balance assessments (Kang, 2015).

The three-dimensional movement and balance stimulation while horseback could have had an influence on the participants' balance improvements. The passive range of motion potentially had an effective sensory input on pelvis mobility, and ultimately refined independent motor output (Quint & Toomey, 1998). Quint and Toomey used a mechanical saddle for assessment of children with cerebral palsy, however, they found riding the moving saddle resulted in significant improvements in range of anterior/posterior tilt. This functional mobility improvement was also seen in a study by

(Diniz et al., 2020) with hippotherapy intervention in an older adult population. They inferred that hippotherapy could have a beneficial impact of daily living activities of older adults due to significant improvements in balance, show through the TUG test and Functional Reach Test. Similarly, the participants in this study achieved improvements in FAB assessments, and results can be attributed to the three-dimensional stimulation of the horse to the human pelvis.

CONCLUSION

The kinematic data of Horse A and B showed a complete flexion and extension in the right and left hip during a normalized stride. The translation of this pattern was visible in the double flexion and extension seen in each rider. This study successfully captured simultaneous horse and rider kinematic patterns, although significant increase or decrease in ranges of motion was discovered. In addition, when assessing FAB exercises that had opportunity for improvement (<4) and summed across all riders and all exercises, there was significant improvements in FAB scores. This concludes that the three-dimensional stimulation of the horse's gait could have resulted in overall improvements in rider balance scores.

CHAPTER THREE

THERAPEUTIC RIDING IMPROVES PYSCHOSOCIAL AND PHYSICAL WELL-BEING IN OLDER ADULTS

INTRODUCTION

Many factors may affect individuals' psychosocial well-being and quality of life as they continue to age. Primarily, lack of social relationships and poor health are two contributors to the decline in life quality over the age of 65 (Bowling et al., 2003) and physical changes in middle and older age adults (Ryff, 1989). Lack of social relationships and maintaining activity can infringe upon a human's psychosocial health as they age, and it is important to focus on those areas while seeking well-being benefits. Frequent and systematic exercise has been found to be effective in combating these negative effects of ageing (Christmas & Andersen, 2000b; Holviala, Sallinen, Kraemer, Alen, & Häkkinen, 2006; Peterson et al., 2010; Steib, Schoene, & Pfeifer, 2010). Even more, social relations in an exercise environment are an integral piece to improving psychosocial well-being and continued satisfaction with life (McAuley et al., 2000).

To combat these potential issues, many have turned towards animal-assisted therapies and activities because animals have been widely accepted as a form of support. A meta-analysis on animal assisted therapies determined its use was effective for the accomplishing desired outcomes (Nimer & Lundahl, 2015). Especially within long-term care facilities, animal-assisted care sessions were considered to have a significant effect on loneliness experienced by the residents (Banks & Banks, 2002).

The use of equine-assisted psychotherapy has emerged as a popular intervention in mental well-being care. Specifically, the intervention treatment can be in the form of equine-assisted psychotherapy (EAP), equine-assisted learning (EAL), equine-facilitated learning (EFL), and equine-assisted activities (EAA) depending on the professional approach (Equine Assisted Growth and Learning Association, n.d.; Professional Association of Therapeutic Horsemanship, n.d.). A meta-analysis performed by Ping-Tzu Lee, Emily Dakin, and Merinda McLure (P. T. Lee, Dakin, & Mclure, 2016) identified an emerging number of studies incorporating EAP practice. Many of these studies found impressive improvements in quality of life and mood (Cerulli et al., 2014; Farias-Tomaszewski, Jenkins, & Keller, 2001; Pretty et al., 2007; White-Lewis, Johnson, Ye, & Russell, 2019) when implementing an equine assisted intervention. Even more, Kathy Lee, et al (K. Lee, Dabelko-Schoeny, Jedlicka, & Burns, 2019) assessed the perceived benefits of EAP from older adult riders, which included positive social stimulation, relaxion in an outdoor environment, and evoking positive memories in addition to the human-animal bonding. However, the field is still developing and more robust, evidence-based studies are needed.

In addition to the psychosocial benefits of equine-assisted interventions, physical benefits also may occur. Horses produce a four-beat gait pattern and their pelvis mimics similar movements patterns to that of a human. While a rider is mounted on the horse, those cyclic movements are directly transferred to the rider, and they are forced to maintain postural balance while moving (Severyn, Luzum, Vernon, Puymbroeck, & DesJardins, 2022). EAA or EAT could act as an alternative to traditional exercise that is used in some rehabilitation programs to improve balance and gait (American Hippotherapy Associations, n.d.; Bronson et al., 2010; De Araújo et al., 2013; Sterba, 2007). The three-dimensional motions of a walking horse provoke full body, cyclic kinematics in a rider that can help develop strength, balance, and coordination (Benda et al., 2003; Janura et al., 2009a). Specifically, in older adults ages 60-84, EAA and EAT, have shown to promote improvements in balance and lower limb strength (Araujo et al., 2011; De Araújo et al., 2013; D. N. Homnick, Henning, Swain, & Homnick, 2013).

Overall, a few studies have examined the use of EAA with older adults. While these studies have found the benefits of animal assisted therapies and the effectiveness of EAA, they have yet to collaboratively explore the physical and psychosocial well-being of EAA interventions in older adults. Therefore, this pilot study was designed to explore the impact of EAA on the psychological and physical well-being of older adults.

RESEARCH METHODS

This qualitative study explored the experiences of older adults who engaged in an 8-week EAA horseback riding program. This study was approved by the local Institutional Review Board (Clemson University) and the Institutional Animal Care and Use Committee (Clemson University).

Participant Recruitment

In order to pilot this intervention, ten participants were recruited from a local lifelong learning center that serves adults 55 and older. Inclusion criteria for this study were that participants, 1) were 55 or older, 2) had no serious surgery that would limit their physical or mental abilities, 3) were available to complete an 8 week study, and 4) had little to moderate riding experience. In addition, interested subjects were asked to rank on a scale of 1-5 their, (1) fear of falling, (2) fear of riding horses, (3) ability to easily walk a quarter mile, (4) ability to do chores around the house, and (5) ability to walk up a flight of stairs. The horseback riding facilities used in this study had uneven terrain and limited assistive equipment to mitigate fear of falling while riding horseback or navigating the arena, therefore participants were chosen within the mild-moderate spectrum. The research team thoroughly assessed answers and asked the remaining 20 respondents to perform a Fullerton Advanced Balance (FAB) assessment. Within the assessment, participants who were unable to perform a task were also excluded from the study. In order to reduce the number of participants, the final 10 were chosen at random among the group of similar responses and FAB assessments.

Intervention

Participants engaged in one 30-minute horseback riding lesson per week for eight weeks, and participated in a semi-structured interview immediately following their eighth lesson. The lessons and interviews were held at the local University Equine Center. The outdoor arena was approximately 100x200 ft area with sand footing. Two lesson instructors were present to assist the rider while mounting and dismounting the horse. Two horses of calm nature who are regularly used for novice riders or other equine assisted activities were used for horseback riding lessons. One female quarter horse (15.3 hands 20 yo) and one male quarter horse (16 hands 19 yo) horses were used with consistent riders throughout the eight weeks.

Data Collection

There were two males and eight females (67.7 ±3.5 yo, range 60-80 yo) chosen for the study. Participants were given anonymous names in the results description of this paper. Participants had minimal surgeries in the past that did not inhibit their current activity levels or motor functions, including three back and two knee surgeries. All participants were involved in a type of physical activity during their week and were considered moderately active. Three subjects had no prior experience on a horse, two had very minimal, and five had moderate experiences horseback riding but had not ridden in at least 12 years. Members of the research team developed semi-structured interview questions as a guide for a post-intervention interview with each participant. All interviews were conducted by one research team member for consistency. Questions focused on the physical and psychological impacts over their eight weeks of lessons and were designed to provide an understanding of the EAA experience. Each question had probes to elicit additional information from participants. Each interview lasted between 10-30 minutes, and was recorded and transcribed verbatim by one of the research team members.

Data Analysis

For the participant’s demographic data, descriptive statistics and frequencies were calculated. For the interview data, the research team utilized directed content analysis (Hsieh & Shannon, 2005) that focused on the constructs of physical and psychological health. The authors first met and examined the text and ascribed pre-determined codes on physical and psychological well-being. Following theme determination, the team members read the interviews a second time and coded participant responses based on the defined themes. After the interviews were individually coded, the team members met and discussed their results. Themes that were different were discussed until a consensus was reached. As a result of this coding, there was a 95% concurrence with the categorization of data. Data that did not fit in one of the predetermined codes was re-analyzed to determine if it was a sub-theme or a new theme. Responses that were coded differently were discussed and successfully re-defined within a consistent theme, resulting in 100% agreement.

RESULTS

From this directed content analysis we were able to identify many themes related to the psychosocial well-being of the participants (as opposed to psychological well-being more broadly). These included quality of life (QoL), motivation, and social comparison, but also an enhanced sense of physical well-being, with subthemes of gait and balance confidence.

Psychosocial Well-Being

The participants in this study described substantial improvements in their psychosocial well-being as a result of participating in the riding lessons. Enhanced psychosocial well-being was described in this study as enhanced QoL (with subthemes of stress relief, positive emotions, self-efficacy, and psychological confidence), motivation, and social comparison.

QoL

Participants described an improved general sense of well-being, or QoL, as a result of participating in the riding lessons. For example, Stella shared that her QoL improved because it was beneficial *"just to be able to get out, to enjoy being out. And um, just a new perspective, and, and just learning something new at my age is always exciting and I feel like that's part of the aging process, so it's great."* Delores described that the riding lessons changed how she felt about herself: "*Well, I feel like I'm younger… Yeah I do, I feel like (pause) uh, I kind of took a step towards being more active."* Enhanced QoL also was demonstrated through participants describing that the riding lessons helped to relieve stress.

Stress relief

Participants in this study described that the focus required by riding was particularly instrumental in reducing stress. Phillip described looking forward to the lessons, and shared *"…I could see [it was] emotionally very relaxing, clear your head, good for your spirit...It's kind of hard to be stressed… it's obviously an environment that, that removes you from stress making environments…I don't have a phone or anything…there's nothing I can do. Abandon yourself to the, to the process. And that's a good thing."* Stella shared this sentiment, and stated *"…So yeah, I think that's a great stress reliever because you got to keep your mind on it. And that's what it's about."*

Positive emotions

Increased QoL was evident through the presence of positive emotions while riding and after riding. Stella described that she experienced positive emotions during the riding lessons: *"I love learning how to do the maneuvers and that was pretty exciting… I figured I'd just be going around and around in a circle. The first time ya'll said something about 'You're going to take her out here and go around this …' I'm like oh yeah right (laughs) and then actually it worked and I was like oh my gosh. And backing up, that was very exciting (laughs)."* Savannah also described how the lessons provided enhanced positive emotions: *"It puts me in a, it would put me in a really good mood and so for that reason I was more relaxed about what was going on around me…So it brought happiness, it brought excitement and something new in my life and um it was fun."*

Self-efficacy

Participants described that the ability to ride a horse improved their beliefs about what they could accomplish, also described as self-efficacy. As Savannah stated, *"It made me feel better about myself that I could actually do this."* Julia elaborated on this in the following exchange by describing how the increased confidence from riding also generalized to her life:

> *"…I came and did this and you know, realized that I could do it. You know, I didn't even know if I could get on a horse, it had been so long. And even though, I had a little difficulty with my hip, um (pauses) um, I think it gave me more confidence."*
>
> Interviewer: *"Good, confidence in just day to day things?"*
>
> Julia: *"Confidence in reminding myself that I can pretty much do anything I try to do."*

Motivation

Participants described different reasons for remaining engaged in the lessons. Some participants described desiring a sense of accomplishment, a passion for riding, and an opportunity for re-engaging in a past leisure activity or learning to ride, while others did it to share the riding experience with a partner or friend. As Stan stated, *"Well I, (pauses) it certainly wasn't on my bucket list or ever occurred to me until I saw the opportunity and I thought, hey that'd be kind of neat because I listen to my wife ... talk about riding horses so much and having a couple of horses growing up, I thought I would uh just do it."* Kathy described *"I look forward to it every week with anticipation and I enjoy it and I come home and I feel like I accomplished something…."*

Some of the study participants described that the motivation for participation in the lessons came from the desire to learn new skills. As Julia stated,

> *"I'm always trying to learn new things, that's why I'm in [lifelong learning program], to take classes and learn things. So, I was just real pleased that each week I came, it kind of motivated me, energy wise for the week…And I, I'm glad that I did learn to do a few things and was able to you know, maneuver the horse properly and she responded very well to me, so that was good."*

Delores elaborated on the style of learning that was beneficial: *"So I liked learning something new. I did like that part, you know, learning something new each time. So, you know that was good the way you guys, kind of built up the course so each time was a little bit different."*

Social Comparison or Riding as Resistance to Social Expectations

Participants described that their experience riding horses surprised their friends and family. Jessica described the benefit from participating as "*Just having a whole different experience that nobody (laughs) I know had! (laughs).*" Stella identified that her friends and family reacted in different manners: *"Um, it was fun talking about it [riding] and they were like 'That's really cool, that's great you learned to do that' and some people were like 'Really? Why did you want to do that?'"* Delores described that her friends were *"very impressed when I say, 'Oh I'm going horseback riding today.'"* Delores further elaborated that riding horses made her feel that she was doing better comparatively than she had previously, and better than her peers or partner. She stated *"I feel like particularly for my peer group, I feel like "Oh, I'm doing just fine!"...So I do feel (pause) so I don't know if it's the horse or what, but I feel (pause) I feel, I truly do feel mentally younger and I do think that part of it is that physically I have more energy…."*

Physical Well-Being

In addition to the improvements in psychosocial well-being, participants described having an increased sense of physical well-being as a result of the riding lessons with the horses. This increased physical well-being took many forms, including increased range of motion, increased strength and flexibility, and reduced pain. For some participants, this improved well-being was global, as Kathy described: *"... it just makes you feel a little more confident I think. Uh doing something for your health."* For other participants, the increases in physical well-being were more specific, such as changes in pain and sensation in the lower body, as described in this exchange:

Stan: *"Uh, I'm pretty sure I mentioned to you both before. Right after, I had a motorcycle accident back in August of 2015 and I have still, uh, a fair amount of nerve and tendon pain on my right lower leg, but after I ride, for some reason, my legs feel wonderful (laughs). So I guess wrapping around a big think horse has some sort of therapeutic value."*

Interviewer: *"When does that pain end up coming back would you say?"*

Stan: *"it depends, I mean uh, I guess it always comes on slowly and I generally notice it going downstairs and down hills but walking back to the car* [after the riding lessons] *I always feel, extra light on my feet (laughs) It's a weird, it's a good sensation."*

The participants also described that physical well-being also emerged as enhanced gait and balance confidence. Both of these subthemes often were described in relation to engaging in leisure activities with less fear or trepidation than prior to the lessons.

Gait Confidence

Improvements in gait (walking) confidence was described by many participants as being a tremendous benefit from participating in the riding lessons. As Delores described,

> *"I just feel when I'm walking around, that I'm more willing to walk faster…That before it was like, you know I always said, 'If our house is on fire, there's no way I'm running out. Because I am not breaking a hip.' Whereas now, with the treadmill, it's like 'Oh, maybe I'll start to run a little bit.' You know, so yeah, I do feel like 'Oh I think I could ... ' um, so I already walk fast, when I do that. So I'm thinking 'Oh, maybe I'll just start to run a little bit and see how that works.'*

Whereas before it was like well there's no way I would ever do that. You know, it's like I'm not taking that risk. So, I do just feel like I'm stronger and more balanced, whether I am or not, I don't know. But I feel like, I feel more confident. To (pause), you know to like do this stuff. You know take hikes, do things like that."

The increased confidence in one's ability to walk also increased the type and amount of physical activities that participants were willing to engage in. Stan shared:

"The leg issue and then the last several weeks I have been um, planting on my property uh, a lot A LOT, I over bought too many plants (laughing) and it's on a steep hillside, and I just noticed maneuvering the hill side, I don't think about it as much with my leg. And I don't know if it has anything to do with Mallorie [the horse] or the riding but uh, it's a positive from where I was a few months ago…it's a very steep property, and walking across the hill where one leg is lower than the other was more difficult than it seems to be now, but yeah."

Balance Confidence

Balance confidence emerged as a belief in one's own balance, which was improved following participating in the riding lessons. For example, when asked about changes in confidence, Savannah shared: *"… I was thinking about this yesterday how I don't even think about the tripping or whatever now as much, and I had gone for a walk in the woods and I was like, you know, carried my little walking stick just waiting to fall but I didn't… So I said well that was pretty good! ..So I think it um has given me confidence."*

DISCUSSION

This pilot study using EAA with older adults demonstrated that older adults perceived substantial benefits in the areas of psychosocial well-being (including QOL, stress relief, positive emotions, self-efficacy, motivation, and social comparison), and physical well-being (including gait and balance confidence). To our knowledge, there is only one other qualitative study that describes the use of Equine Assisted Psychotherapy (EAP) with older adults (K. Lee et al., 2019). Although the actual intervention (EAP) is different than the current study (EAA), the use of equine-related activities with older adults is similar to the current study. Lee and colleagues found that older adults reported experiencing positive memories, and enjoyment in being outdoors during EAP, and the older adults in the current study experienced similar benefits. In an observational study with six older adults with dementia, Fields, Bruemmer, Gloeckner, & Wood (Fields, Bruemmer, Gloeckner, & Wood, 2018) found that an equine assisted activities program (EAAP) resulted in enhanced emotional well-being. While these studies reflect some similar findings to the current study, this is the first study to identify changes in QoL, stress relief, positive emotions, self-efficacy, motivation, and social comparison in the older adult population following an EAA intervention.

Studies of EAA with older adults have focused primarily on the physical outcomes from the intervention. This pilot study supports the work by Homnick and colleagues (D. N. Homnick et al., 2013; T. D. Homnick, Henning, Swain, & Homnick, 2015) which described therapeutic riding activities as safe for older adults with mild to moderate balance deficits. Changes in balance were also perceived by the participants in the current study which is consistent with the findings by White-Lewis et al (White-

Lewis et al., 2019) who found balance improved in 20 adults and older adults with arthritis, and with Homnick et al (D. N. Homnick et al., 2013; T. D. Homnick et al., 2015), who found that balance improved in a general population of older adults following EAA. Other studies have noted enhanced postural stability and dynamic control (Wehofer, Goodson, & Shurtleff, 2013), and improvements in pain and range of motion (White-Lewis et al., 2019). The current study also found improvements in the perception of pain and range of motion. Two studies with 9 older adults with Alzheimer's disease indicated improved balance and functional capacity following an EAT intervention (Araujo et al., 2011; Borges De Araujo et al., 2019). Finally, the participants in the current study reported that they experienced improvements in both gait and balance confidence, which supports the case study of a 76-year woman who reported a reduced fear of falling following a six-week EAA and EAAT intervention Wehofer, Goodson, & Shurtleff (Wehofer et al., 2013).

Collectively, this study adds to the literature by identifying novel outcomes from an EAA intervention (QoL, stress relief, positive emotions, self-efficacy, motivation, social comparison, gait confidence, and balance confidence) that were perceived by the older adults in this study. These perceived outcomes have implications for future research, including the quantitative measurement of these outcomes, and an in-depth exploration of how these outcomes influence the global well-being of the older adult.

While this study was a pilot study, there are a few limitations which should be noted. Although we were not certain about what the outcomes could have been, balance measures could have been implemented to determine if the perceived changes matched the measured changes. Furthermore, a small sample and qualitative data reduce generalizability, but provides additional information to support additional studies.

Implications for Recreational Therapy Practice

To our knowledge, there are four published papers that examine EAA or EAT within the context of recreational therapy. In a literature review, Duffy (Duffy, 2018) reported that in recreational therapy practice, QoL can be enhanced through EAL. Corring, Johnston, and Rudnick (Corring, Gath, & Rudnick, 2010) found that in adults with schizophrenia, enhanced enjoyment, bonding with the horses, and increased self-confidence occurred. Goodwin, Hawkins, Townsend, Van Puymbroeck, and Lewis (Goodwin, MS, CTRS, TRS, Hawkins, PhD, CTRS, LRT, Townsend, PhD, CTRS, Van Puymbroeck, PhD, CTRS, FDRT, & Lewis, PhD, CTRS, 2016) found enhanced self-efficacy in children with Autism Spectrum Disorders (ASD) following EAA, while Hawkins, Ryan, Cory, & Donaldson, (Hawkins, Ryan, Cory, & Donaldson, 2014) found that body coordination, strength, agility, and gross motor skills improved for two children with ASD. Together with the results of the current study, these studies indicate that recreational therapists may find benefit in using EAA or equine-related activities as a therapeutic intervention for clients/patients of a variety of ages and diagnostic groups.

CONCLUSION

Participants in this study expressed perceived substantial benefits in psychosocial well-being (QoL, stress relief, positive emotions, self-efficacy, motivation, social comparison, gait confidence, and balance confidence). In conjunction with quantitative studies that assess range of motion, participants of this study also perceived improved range of motion. The implication of these results could benefit the exploration of the perceived effect of EAA on older adults.

CHAPTER FOUR

COMPARISON OF TRUNK AND LOWER LIMB MUSCLE ACTIVITY WHILE HORSEBACK RIDING AND DURING HEALTHY HUMAN GAIT

INTRODUCTION

Hippotherapy is a physical rehabilitation practice that uses horseback riding as a means of providing motor and sensory input to promote functional outcomes (American Hippotherapy Associations, n.d.). Hippotherapy enables the improvement of muscle strength, balance, and coordination, as patients' muscular activation patterns are forced to adapt to compensate for the movement of the horse, eliciting a full body muscular response (Angoules & Kapari, n.d.; De Araújo et al., 2013). This method has been investigated as a rehabilitation tool for those with cerebral palsy, multiple sclerosis, stroke-induced hemiparesis, and spinal cord injury, among other conditions (Beinotti et al., 2010; De Araújo et al., 2013; Koca & Ataseven, 2015). Hippotherapy aligns with the goals of traditional physical therapy by providing repeated visual, vestibular, and somatosensory stimulation (Araujo et al., 2011). Additionally, hippotherapy offers unique benefits, including induction of motion patterns and muscular activation that cannot be produced independently (Beinotti et al., 2010). This may be especially useful for those recovering from neurological, physical or cognitive trauma.

Previous studies show promising evidence of the effectiveness of hippotherapy, including documented improvements in physical outcomes as well as shortened recovery times (De Araújo et al., 2013; Y. N. Kim & Lee, 2015; Koca & Ataseven, 2015). In studies on hippotherapy and balance improvement, hippotherapy has been shown to decrease sway in gait, to improve lower limb strength and balance, and to decrease fall risk in healthy, elderly subjects (Araujo et al., 2011; De Araújo et al., 2013; Y. N. Kim & Lee, 2015; M. Lee et al., 2014; Severyn et al., 2022). These positive findings have extended to targeted rehabilitation efforts, as a 2010 study found that hippotherapy brought the gait of hemiparetic post-stroke patients closer to normality than traditional rehabilitation methods (Beinotti et al., 2010).

Hypothesized similarities between the biomechanics of horseback riding and normal gait have been a key motivation for the use of hippotherapy. The rationale behind this theory is that the motion of a horse stimulates movement patterns and muscular activation that mimic normal human gait (Biery, 1985; Biery & Kauffman, 1989; Garner & Rigby, 2015b). A 2015 study provided evidence in support of this theory, finding many similarities between pelvic motion patterns in gait and riding in healthy children (Garner & Rigby, 2015b). In addition to kinematic research, the muscle activation patterns experienced during horseback riding have been explored in a few recent studies. Kim and Lee (S. G. Kim & Lee, 2015) found that muscle activation increased significantly after 8 weeks of intervention in an elderly cohort, although this study employed a riding simulator. Another study (Elmeua González & Šarabon, 2020) examined activation of core, upper body and lower body muscles in both recreational and professional horseback riders using surface electromyography (EMG). This group successfully characterized the muscle activation patterns of horseback riding, and discovered greater control of muscle activation by professional riders. However, this data must be interpreted with hesitancy when considering rehabilitation implications, as even the recreational riders had at least 17 years of experience (Elmeua González & Šarabon, 2020). EMG patterns in targeted hippotherapy have also been explored, with one group finding an increase in lower limb muscle activity in cerebral palsy patients after 10 sessions (Ribeiro et al., 2019). While this work has begun to define muscle activity as it relates to hippotherapy, further exploration of this relationship is necessary. The comparison riding data to that of standard gait is especially relevant to the rehabilitation of conditions that prioritize gait correction, such as hemiparetic stroke rehabilitation (Beinotti et al., 2010). This integration of data has the potential to deepen understanding of human biomechanics and bodily responses while horseback riding.

The muscle activation patterns of normal human gait have been extensively characterized. Winter and Yack (Winter & Yack, 1987) published detailed patterns of EMG activity during the gait cycle in healthy subjects. A primary finding of this study is that the more proximal muscles involved in gait display the most stride-to-stride variability, and also have lower magnitudes of activation in comparison to distal muscles of the lower limb. It is also noted that these proximal muscles are those that are most involved in the maintenance of balance and posture (Winter & Yack, 1987). The relevance of normal muscle activation patterns in gait rehabilitation has been widely researched, and varies considerably between patient populations. For example, changes in activation patterns have been linked to improvement in gait patterns in patients with Parkinson’s disease, incomplete spinal cord injury, and hemophilic arthropathy(Cruz-Montecinos, Pérez-Alenda, Querol, Cerda, & Maas, 2020; Govil & Noohu, 2013; Rizzone, Ferrarin, Lanotte, Lopiano, & Carpinella, 2017). On the other hand, improvements in gait have not been linked to changes in muscle activation in hemiparetic stroke patients (Buurke et al., 2008; Den Otter, Geurts, De Haart, Mulder, & Duysens, 2005). However, this finding may reflect a poor ability to improve muscle activation in these patients with traditional strategies, rather than a lack of benefit from this potential outcome.

The purpose of this study was to compare the muscle activation patterns and magnitudes of horseback riding and normal gait in muscles relevant to gait and balance. In this study, we hypothesize that activation of the rectus abdominis, erector spinae, rectus femoris, and biceps femoris muscles will be similar in both frequency and magnitude across gait and riding. These muscle groups were selected for their important roles in gait and/or balance and stability (Winter & Yack, 1987). Large, superficially located muscles were also prioritized to ensure reliability of surface EMG recordings. To explore this hypothesis, surface EMG recordings were obtained during walking gait and horseback riding in healthy subjects. From this work, we aim to improve the current understanding of horseback riding biomechanics for future implementation of hippotherapy.

RESEARCH METHODS

EMG data was collected in walking and riding experiments with two control groups: experienced and novice horseback riders. Surface EMG was used to measure muscle activation throughout the gait cycle of walking and horseback riding. The same subjects were used for both walking and riding trials for proper comparison of results. This study was approved by the Clemson University Institutional Review Board (IRB) and by the Institutional Animal Care and Use Committee (IACUAC).

Participants

Nine healthy female subjects (age 18 - 22) were recruited to participate in this study. These subjects were divided into two groups, “novice” and “experienced” riders. The “novice” rider group (n=5) had self-reported less than 10 hours experience, and the “experienced” group (n=4) had self-reported as having greater than 10 hours of significant riding experience or previous lessons. Exclusion criteria included any type of back, hip, or other surgeries or ailments that could skew the results. Subjects were also excluded from the study if they could not walk over rough terrain, required any device to help them walk (such as a cane, wheelchair, crutches, or walker), or had more than a moderate fear of falling and riding a horse. Volunteers were not compensated for their time in the study, with the only cited benefits being the opportunity to ride a horse.

Experimental Setup

The horse, saddle, and arena footing were constant throughout all riding trials to eliminate variability. The arena was dimensionally 30x60 meters with the surface consisting of 63.5mm river sand on top of a clay base. The horse used was a 21 year old quarter horse and 15.3 hands. An English saddle was used for all riders. On the day of testing, each participant arrived, reviewed the study protocol and signed the participant consent form.

EMG Experimental protocol

Each participant was instrumented with eight surface electromyography (EMG) sensors (Biometrics, LTD, United Kingdom) with a fixed electrode distance of 20 mm to quantify the muscle activation patterns of human gait and horseback riding. The sensors were placed on the following muscle groups bilaterally: rectus femoris, biceps femoris, rectus abdominis, and lumbar erector spinae (Figure 19). Each subject's skin was abraded with 70% isopropyl alcohol (IPA) pre-saturated wipes prior to sensor placement, and SENIAM guidelines for sensor placement were followed for rectus femoris, biceps femoris, and erector spinae (Stegeman & Hermens, n.d.). Rectus abdominis sensory placement was adapted from a 2013 study(Cuesta-Vargas & González-Sánchez, 2013)

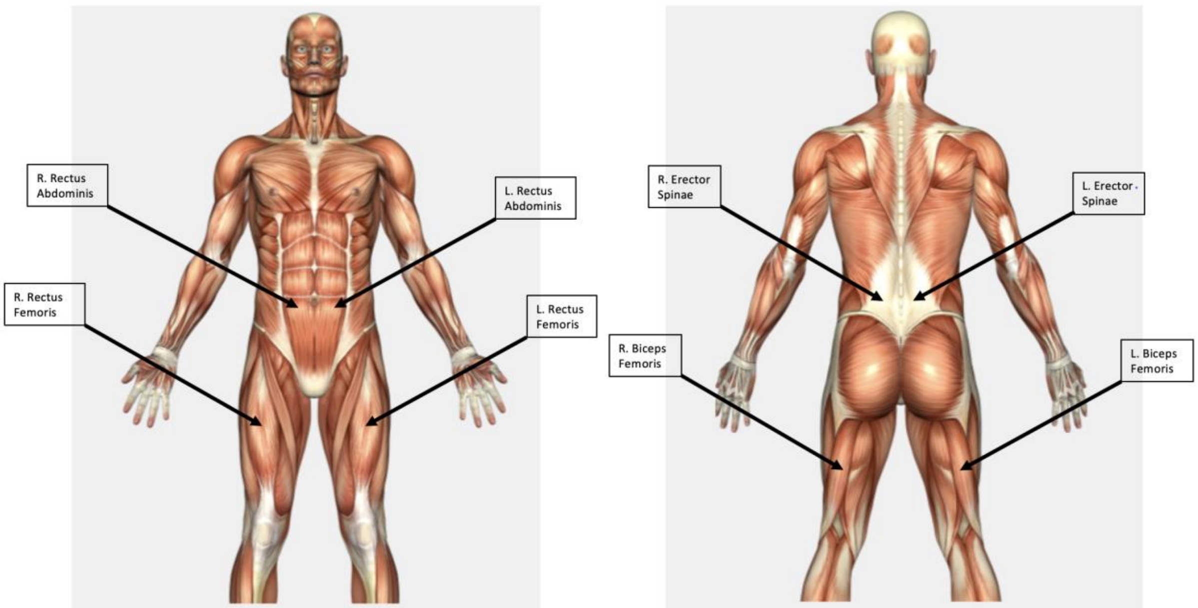

Figure 19. Selected muscles for EMG analysis. Anterior muscles (bilateral rectus abdominis and rectus femoris) are labeled on the left. Posterior muscles (bilateral erector spinae and biceps femoris) are labeled on the right.

Before conducting walking and riding trials, maximum isometric muscle contractions were obtained for data normalization purposes. The procedures for these maximum contractions were adapted from a 2008 study on muscle activation in cycling and from Daniels and Worthingham's muscle testing guide (Helen Hislop, Avers, & Merybeth Browns, 2013). For maximum contraction of the rectus femoris, the subjects, sitting with hips and knees at 90-degree angles, were instructed to attempt to extend their leg at the knee while matched resistance was provided by resistance bands. For the biceps femoris the same procedure was followed, but the subject attempted to flex their leg at the knee joint with applied resistance from the opposite direction. For the rectus abdominis, the subject was instructed to attempt a sit up from a supine position, and a research team member matched the resistance of the subject once their shoulder blades became elevated by exerting bilateral force on the anterior portion of their shoulders. To obtain a maximum contraction of the lumbar erector spinae, the subject attempted to raise their chest off the ground while lying with matched resistance from a research team member, who exerted force on their shoulder blades once their chest was slightly raised.

Horseback riding trials

For riding data collection, the horse and rider were led by rein by an experienced equestrian at a walk. The walking circuit consisted of 3 passes through an oval volume measuring 40x20 feet. In this space video was collected to synchronize the EMG signals to the kinematic sequence of the horse. The video capture allowed for stride separation and normalization of the data collected. In this case, a stride was considered to be the time between two front left fetlock extensions, and one stride was analyzed from each pass through the arena.

Walking Trials

To obtain gait data, the subject was instructed to walk on a treadmill at 2.5 mph for approximately 30 seconds. Video analysis was used to isolate three individual strides, which were defined as the time between 2 left heel strikes. A hand-held lab camera was used to capture the strides for stride separation and normalization.

Data Processing

EMGTools software (Nikolic & Krarup, 2011) was used to rectify and filter the raw EMG data (Bandpass, 50 - 500 Hz), as well as to normalize all data points as a percentage of the maximum contraction. After this, the data was normalized to 0 - 100% of the gait cycle using a custom MATLAB script to ensure there are equal distribution of discrete data points across all strides, with an averaged data point at each percentage point (i.e. 101 data points for 0 - 100%). This data was then averaged for each of the three horseback-riding strides and walking strides to create one characteristic pattern for each muscle for both walking and riding for each subject. Finally, the average and maximum muscle activation for each averaged waveform were recorded. Each average waveform was also graphed to enable qualitative comparison of muscle activity patterns. Waveforms were not created and analyzed for each muscle group for all participants, as several sensors became disconnected from the skin during horseback riding trials.

Data Analysis

Statistical analysis was performed to compare the muscle contractions between novice riders and experienced riders within the same muscle group. Average muscle contractions for walking and riding were compared for novice and experienced riders separately using a paired t-test. Maximum muscle contractions for riding and walking were compared for both novice and experienced riders separately using a paired t-test. The average and maximum muscle contractions during riding were compared between novice and experienced riders using two-sample t-tests. Statistical analysis was not performed for the left biceps femoris for both control groups, as well as the right biceps femoris for the novice group, due to previously mentioned instrumentation disruptions, which led to an insignificant sample size. A significance level of 0.05 was used for all statistical tests.

RESULTS

A sample EMG waveform plot is shown in Figure 20. The average and maximum values for each individual EMG waveform for all muscle groups were averaged within participant groups and are represented in Figure 21 and Figure 22. The novice biceps femoris data is not a true average across subjects, but rather the results from the one rider in this group with successful data collection. A summary of statistical tests performed is shown in Table 10. Statistical analysis of average and maximum excitations revealed significant differences in only one muscle group – the erector spinae – between riding and gait. This difference was only present in the experienced group, with greater activity while riding. Additionally, only maximal rectus abdominis activity differed significantly in riding between the novice and experienced groups. Average and maximal muscle activation for the remaining muscles did not differ significantly between riding and gait, as well as between groups in riding.

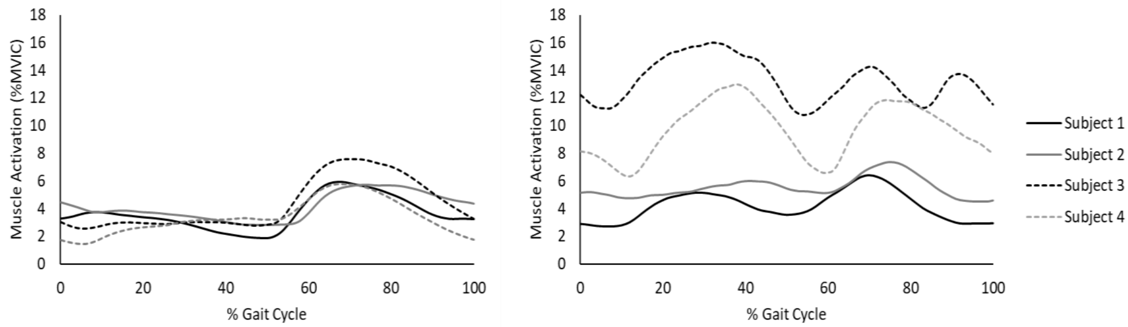

Figure 20. Sample EMG waveform plot demonstrating excitation of the left erector spinae in the experienced group while walking (left) and horseback riding (right). MVIC = Maximum Voluntary Isometric Contraction.

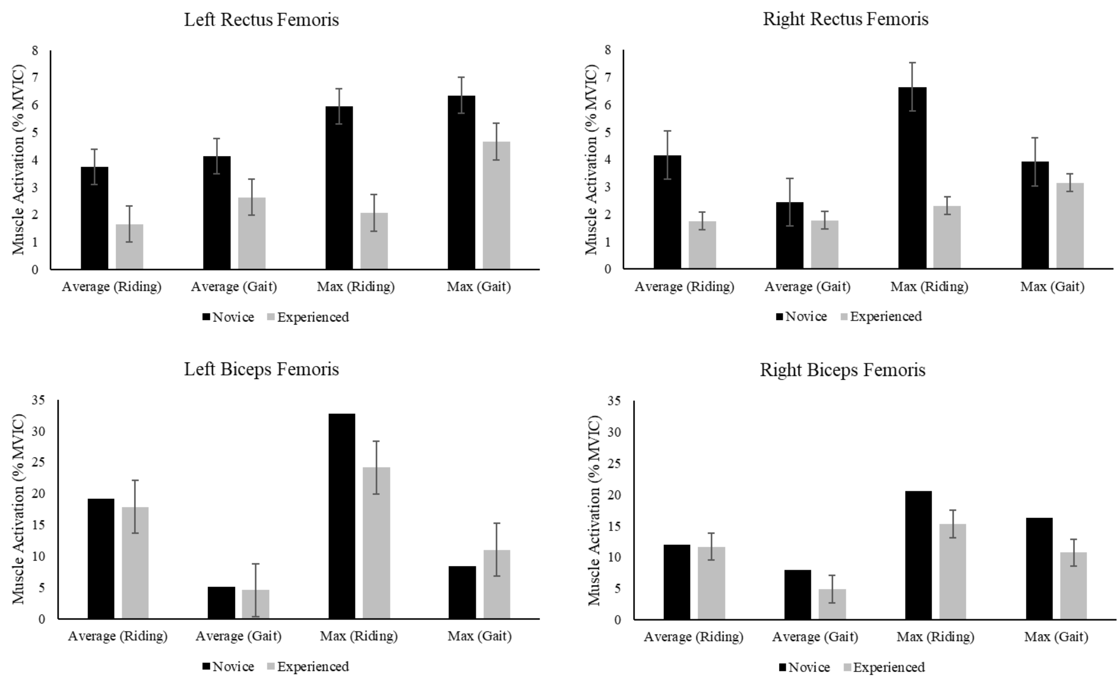

Figure 21. Summary of lower limb muscle activation in riding and gait. Biceps femoris data was collected on just one novice rider. MVIC = Maximum Voluntary Isometric Contraction.

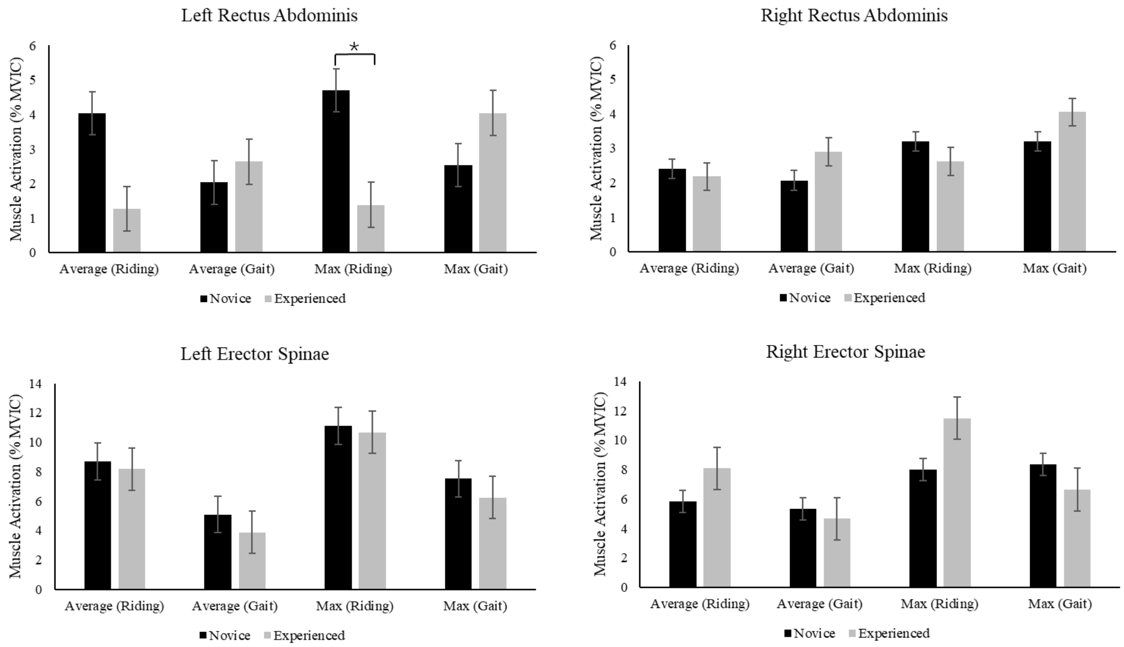

Figure 22. Summary of core muscle activation in riding and gait. MVIC = Maximum Voluntary Isometric Contraction. *statistically significant

Table 10. EMG statistical analysis of novice and experienced group comparisons across muscles (RF = Rectus Femoris, BF = Biceps Femoris, RA = Rectus abdominis, ES = Erector Spinae). Biceps femoris data could not be analyzed in most tests due to insufficient sample size. ****statistically significant*

Comparison	**Left RF**	**Left RA**	**Left ES**	**Right RF**	**Right BF**	**Right RA**	**Right ES**
Novice Average - Walking vs. Riding	0.54	0.12	0.14	0.09	N/A	0.73	0.72
Experienced Average - Walking vs. Riding	0.37	0.20	0.12	0.96	0.08	0.77	0.02*
Novice Maximum - Walking vs. Riding	0.74	0.10	0.19	0.18	N/A	1.00	0.21
Experienced Maximum - Walking vs Riding	0.22	0.21	0.11	0.40	0.38	0.70	0.03*
Riding Average - Novice vs. Experienced	0.17	0.06	0.86	0.24	N/A	0.85	0.31
Riding Maximum - Novice vs. Experienced	0.14	0.04*	0.91	0.20	N/A	0.69	0.25

DISCUSSION

Muscle Activation Magnitudes

Similarities in motion between horseback riding and gait may have useful implications for physical rehabilitation. In this study, few significant differences existed between the muscle activity of walking and riding in both experienced and novice horseback riders. This result was expected, as all riders were healthy and had no known gait abnormalities. In the only observed significant difference between walking and riding, the right erector spinae had greater average and maximum contractions in the experienced control group while riding in comparison to walking. This finding suggests the erector spinae may be more active in riding than in walking as experience grows. Or, the horse could be eliciting asymmetrical movement on the riders due to a gait abnormality that is not visible by the naked eye.

Additionally, the left rectus abdominis of novice riders exhibited significantly greater maximum excitation while riding in comparison to experienced riders. A possible explanation for this result is that novice riders may exhibit more muscle tension in response to a lack of balance on the horse. It has been found that an increase in this muscle tension can, consequently, increase intra-abdominal pressure and require a stabilization of the trunk by stiffening in the spinal column (Cresswell, Ingimar, & Oddsson, n.d.; Tesh, Dunn, & Evans, 1987). This stiffening of the spinal column could manifest itself in the activation of the erector spinae. Therefore, the significant increase in activation among the erector spinae and rectus abdominus could have been a result of this relationship. However, a larger sample size is needed to further investigate.

The rider is required to activate their trunk muscles in order to maintain postural control and follow the horse's movement by swinging their pelvis forward and up (K Terada, Mullineaux, Lanovaz, Kato, & Clayton, 2004). Asymmetry in human gait is largely focused on during rehabilitation for a number of mobility impairments. However, promoting muscle activation in gait is difficult without the stimulation of an outside source, such as a horse or horseback riding simulator. This postural control relationship between the rider and horse or horse simulator has been shown to greatly decrease trunk asymmetry in riders (M. J. Kim, Kim, Oh, & Yoon, 2018; S. G. Kim & Lee, 2015; McGibbon, Benda, Duncan, & Silkwood-Sherer, 2009). Therefore, if similar muscle groups to human gait can be activated while horseback riding, it can promote muscular asymmetry for riders.

All other muscles in both groups exhibited no significant differences between walking and riding. Similar muscle activity in the selected muscles, which are relevant for both gait and balance, indicates that horseback riding may stimulate these muscles in a way that could be useful in the physical rehabilitation of many conditions, providing promising evidence in support of hippotherapy use (Winter & Yack, 1987).

Muscle Activation Waveforms

Qualitative comparisons of EMG waveforms offer additional insights into the activity of the muscles analyzed in this study in riding and gait. The rectus femoris was activated more predictably in gait, as peaks on the left leg tended to appear towards the beginning of the gait cycle, with peaks in the right rectus femoris closer to the end. This pattern is consistent with previously reported findings from normal gait (Winter & Yack, 1987). Activation patterns of these muscles were more variable in riding than in gait for both groups, suggesting that horseback riding does not elicit the predictable activation patterns of gait in the proximal muscles of the lower limbs. Biceps femoris waveforms could not be analyzed to the extent of other muscle groups due to low sample size. Although data from the biceps femoris was only successfully collected in one novice rider, similar trends were observed in the waveforms of this subject and the experienced riders in comparison to the rectus femoris. In both muscles, activation was more predictable in gait, but riding elicited similar magnitudes, often with multiple distinct peaks and valleys in activity. While the importance of activation patterns vs. magnitudes towards positive outcomes is unknown, the reported outcomes of gait correction and balance improvements in previous hippotherapy research suggest that this similarity in magnitude may have clinical significance (De Araújo et al., 2013; Y. N. Kim & Lee, 2015; C. W. Lee, Kim, & Yong, 2014; Severyn et al., 2022).

In the rectus abdominis muscle, it appears that experienced riders exhibit a more constant muscle tone throughout the horse gait cycle, while novice riders may be prone to more spikes in activation. This suggests that novice riders are more unstable than experienced riders, which has been observed in previous studies (Kayo Terada, 2000). Previous research has found a two peak pattern in rectus abdominis activity while riding, and these results suggest this response may be increased in novice riders, although some only exhibited a singular peak (K Terada et al., 2004). While novice riders did exhibit these spikes in activation, muscle tone was fairly constant with no more than a 2% range in activation throughout the gait cycle. Similar constant rectus abdominis tone was commonly observed in gait with a few deviations. Therefore, without any significant differences in activation between walking and riding, it seems that these activities may elicit similar activation patterns.

The similarities between walking and riding were most evident in the erector spinae waveform graphs, especially in experienced riders. A two-peak EMG waveform is expected in this muscle during gait, which was observed in most subjects (Winter & Yack, 1987). This waveform also appeared in the riding waveform of three experienced riders. While this was not evident in novice riders, many of these subjects experienced similar or greater erector spinae activation while riding. A 2000 study found a similar lack of consistent activation in novice horseback riding erector spinae data in comparison to experienced subjects, although this study compared EMG frequency instead of magnitude (Kayo Terada, 2000). Several subjects from both control groups exhibit a peak in activation near 70% of the gait cycle in both riding and waking. This data provides evidence supporting the similarities in erector spinae activation patterns between walking and riding, and shows that this similarity may become stronger with riding experience. This is again supported by previous findings of novice rider instability, which dampens the ability of the rider to move in a rhythmic manner (Kayo Terada, 2000).

Variability

Data was moderately variable between walking and riding, and in many cases between subjects, even within the same control groups. Variability in this data was expected, especially in horseback riding. While muscle activation in gait is somewhat predictable, different subjects can exhibit nearly identical kinematics with a wide range of combinations of muscle activity (Winter & Yack, 1987). This even holds true in muscles involved directly in leg motion, such as the biceps femoris and rectus femoris, and can manifest as differences in contraction magnitude and excitation patterns (Winter & Yack, 1987). Therefore, with this expected variation in gait muscle activity, it logically follows that riding would elicit even more variable data, especially in subjects with little to no experience. Our group is currently conducting research that integrates EMG recordings with 3-dimensional motion capture, which may be useful in providing additional insight into the kinematics associated with muscle activation in riding.

Limitations

Some data sets were excluded due to instrumentation and data loss issues (such as EMG's coming loose during testing). This led to the exclusion of data from certain muscle groups for two riders from each of the novice and experienced groups, especially for the biceps femoris. Contact of the saddle with the posterior thigh often caused the biceps femoris sensor to slip, disrupting data collection. Due to these issues and the initially small sample size of this study, future work with additional subjects will be needed to expand upon these results.

The methods of MVIC measurement may have also contributed to variability seen in muscle activation magnitudes. Some subjects were able to overcome the force provided by the elastic resistance bands, resulting in a contraction that was not truly isometric. Additionally, there is inherent variability in methods requiring subject compliance, as it cannot be guaranteed that each subject exerted maximal effort. Therefore, the magnitude of normalized muscle excitation may not be entirely representative of a fraction of maximal contraction, especially in the muscles that were tested with resistance bands (rectus femoris and biceps femoris).

CONCLUSION

These results suggest the activity of core and proximal lower limb muscles involved in walking and horseback riding may be similar. This is very promising for future implementation of hippotherapy as a rehabilitation tool, especially in patient populations with a significant need for gait correction. Standing alone, this research provides unprecedented insight to the similarities between muscle excitation in normal human gait and while riding horseback. Future work involving integration with 3-D motion capture data may provide a greater understanding of the relationship between the biomechanics of walking and riding. Ultimately, the initial data points towards the use of hippotherapy as a tool for gait correction, and the methods used in this study can be applied with few deviations for future patient research.

CHAPTER FIVE

INFLEUNCE OF HORSE'S LUBMOSACRAL JOINT ON THE KINEMATIC AND JOINT MOMENT RESPONSES IN THE RIDER

INTRODUCTION

Pelvic-core mobility of Equine Assisted Activities and Therapies

Biomechanically, a horse's back can be thought of as a movement platform, that induces multi-planar motion about a roll, pitch, and yaw axis (Byström, Rhodin, von Peinen, Weishaupt, & Roepstorff, 2009b; Clayton & Hobbs, 2017). As a result, a rider's pelvis reacts to this movement and undergoes two pitching cycles, a single cycle of yaw and roll with each stride of the horse (Byström, Rhodin, Peinen, et al., 2009). Equine assisted activities and therapies (EAAT) allow the rider to engage (both kinematically and muscularly) with most pelvic movement patterns associated with normal human gait (Garner & Rigby, 2015b). This movement pattern is influenced primarily by the horse's hind legs. For example, as the left hindlimb lifts off the ground, the horse's trunk and rider's pelvis roll towards that hind limb. When the left hindlimb hits the ground again, the rider's pelvis will pitch posterior until the left forelimb contacts the ground, to which it will pitch anterior. This cycle repeats when the opposite hindlimb and forelimb contacts the ground (Byström, Rhodin, von Peinen, et al., 2009b). This movement by the rider is not unlike the actual walking motion of a normal person, but specific studies to quantify this movement, and its potential benefits to stroke recovery patients have not been well quantified. Surprisingly, the psychological benefits of horse-human interaction have been well documented. In fact, this psychological bonding often results in enhanced patient compliance with treatment, and increased overall patient quality of life outcomes.

Equine Assisted Activities and Therapies as an alternative form of physical rehabilitation

EAAT improves pelvic range of motion and balance as well as the ability to increase muscle strength in the pelvic region (Garner & Rigby, 2015b). A 2014 study found similar balance improvements between control groups that underwent treadmill and EAAT. However, these control groups had significant difference in gait velocity and symmetry outcomes, with the EAAT control group experiencing greater improvement (C. W. Lee et al., 2014). These results suggest that EAAT may have the potential to improve patient gait and balance more effectively than traditional methods. As discussed previously, this is significant due to the importance of gait improvements in stroke patients.

A study by Benda et al. (Benda et al., 2003) assessed muscle activity via remote surface EMG sensors in hippotherapy intervention on children with cerebral palsy (n=15). The control group was tasked with eight minutes astride a barrel and the intervention group received eight minutes of hippotherapy, muscle excitation was observed immediately pre and post the sessions. Significant improvements in muscle symmetry were found in the intervention group, and no significant improvements were found in the barrel group.

Current mobility assessment in Equine Assisted Activities and Therapies

EAAT aligns with the goals of traditional physical therapy strategies by providing repeated visual, vestibular, and somatosensory stimulation, accompanied by a few unique benefits (Araujo et al., 2011). Most relevant of these unique benefits is that horseback riding allows patients to experience motion patterns, muscular activation, and muscular stimulation that cannot be produced independently (Beinotti et al., 2010), especially for those recovering from neurological, physical and/or cognitive trauma.

A 2015 study found that horseback riding emulates many, but not all aspects of pelvic motion in healthy children (Garner & Rigby, 2015a). Further quantitative investigation is required to expand upon the results of this study, and to further explore the relationship between human pelvic kinematics, core muscular activation and riding-induced perturbation. This integration of data has the potential to deepen understanding of human biomechanics and bodily responses while horseback riding. The comparison of these results to that of standard gait is especially relevant to the rehabilitation of conditions that prioritize gait correction, such as hemiparetic stroke rehabilitation (Beinotti et al., 2010).

Current gait research analyzes various EMG patterns for various muscle groups to determine effects of various diseases have on one's gait cycle. Analyzing EMG patterns is useful in determining if various exercises or motions can improve one's gait as seen in the 2015 study that proved that trunk exercises performed on unstable surfaces improved the muscle activation of the trunk muscles in stroke patients (Y. N. Kim & Lee, 2015). Muscle activity patterns are also observed to determine changes in gait patterns after a serious medical event such as a hemophilic arthropathy (Cruz-Montecinos et al., 2020). Understanding the effects of various medical conditions and exercises have on effecting a human gait can result in developing better ways to improve the human gait. There are various studies analyzing the effects of equine assisted therapy on muscle activation patterns (M. J. Kim et al., 2018), yet there are not many significant studies analyzing the effects hippotherapy has with stoke patients.

Another study assessed the clinical gait and balance improvements between post stroke control group and mechanical horseback riding intervention. Participants were recruited (n=37) and divided into traditional physiotherapy and mechanical horseback riding and physiotherapy rehabilitation methods. The study utilized Functional Ambulation Category (FAC), the gait portion of the Performance Oriented Mobility Assessment (G-POMA), Berg Balance Scale (BBS), and the balance part of the Performance Oriented Mobility Assessment (B-POMA). No significant difference was found between groups in initial baseline values, however, between the baseline and post-trial assessment the intervention showed significant improvement on the BBS, B-POMA, and gait parameters (Han et al., 2012).

However, most of these studies lack robust quantitative assessments of 3-dimensional horse and patient movement during EAAT. As stated previously, there are also many unknowns in approaching stroke rehabilitation, specifically including the intensity of training. While current evidence supports the positive influence of hippotherapy on balance, the biomechanics of the rider and horse during horseback riding that produces these improvements is largely unexplored. While other studies (Beinotti et al., 2010; Byström, Rhodin, von Peinen, et al., 2009b; Garner & Rigby, 2015b; Münz et al., 2014a) all used motion capture technologies to track rider kinematics, they did not investigate the relationship between horse and rider movement, and discuss these kinematic interventions with respect a larger health related outcome such as balance. By defining the kinematic relationship between the horse and rider, methodologies and models can be developed and implemented to specific patient populations. Therefore, the purpose of this study is to develop a quantified representation of the horse and rider relationship, and further investigate the correlation between the horse's pelvic movements as a linkage system, and its resulting effect on the rider's pelvic reactions.

RESEARCH METHODS

Study Design and Patient Recruitment

Using Qualisys motion capture technologies and Visual 3D model generation and analysis, horse and rider model was successfully created for kinematic and kinetic evaluation.

Desired reflective marker locations were chosen based on anatomical preferences for model generation (Figure 23). Tall segments were created within Visual3D as cylinder segment shapes. The rider's head, thorax, and arms were modeled in reference to the RAB Upper Extremity Model (Rab et al., 2002) with some variations. Markers along the shoulders, sternum, thoracic spine, wrist, and hands were mimicked of the RAB Upper Extremity Model. However, some modifications were made to be adaptable to a horseback rider. To accommodate the helmet on the rider's head, the markers were placed on the anterior cranial, superior cranial, and posterior cranial along a sagittal plane line. The CODA pelvis was chosen from the Conventional Gait Model (C-motion, n.d.) using the right and left ASIS and PSIS. From the same tutorial, the lower limbs were mimicked similarly with minor modifications.

Similar methods were followed to create the skeletal model of the horse. The horse's head, thorax, and pelvis were modelled as cones and all other remaining segments as cylinders. All bone visuals were generated using Visual3D graphics (Figure 23).

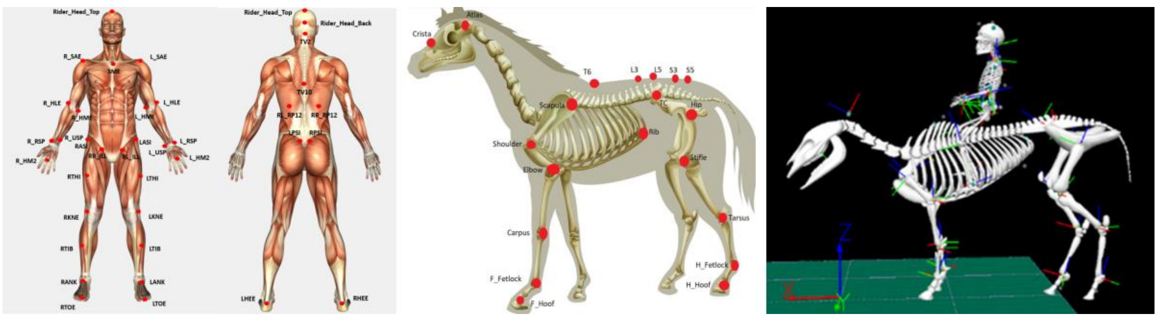

Figure 23. Anatomical joint markers for rider and horse, and resulting 3D model within Visual 3D

Table 11. Human body segment geometry and defining points to generate horseback rider model. Right and left limbs were labelled as "R" and "L" abbreviations.

Name	Shape	Location	Defining Point 1	Defining Point 2
Pelvis	Cylinder	Right	RIDER_RASI	RIDER_RPSI
		Left	RIDER_LASI	RIDER_LPSI
Right/Left Thigh	Cylinder	Proximal	RIGHT_HIP/LEFT_HIP	-
		Distal	RIDER_RKNE/RIDER_LKNE	-
		Extra Target	Lateral, RIDER_RTHI/RIDER_LTHI	-
R/L Shank	Cylinder	Proximal	CODA_RIGHT/LEFT_KNEE	Radius: 0.5*Right/Left_Knee_Width+Marker_Radius
		Distal	RIDER_RANK/RIDER_LANK	Radius: 0.5*Right/Left_Ankle_Width+Marker_Radius
		Extra Target	Medial, (R/L)_FRAMECENTER	-
R/L Foot	Cone	Proximal	RIDER_RANK/RIDER_LANK	RIGHT_ANKLE/LEFT_ANKLE
		Distal	RIDER_RTOE/RIDER_LTOE	Radius: 0.06 m
Thorax	Cylinder	Proximal	RIGHT_ILIAC	LEFT_ILIAC
		Distal	RIDER_R_SAE	RIDER_L_SAE
Head	Ellipsoid	Proximal	C7_STRN	-
		Distal	RIGHT_DISTAL_SKULL	LEFT_DISTAL_SKULL
R/L Upper	Cylinder	Proximal	RIDER_R/L_SAE	Radius: 0.055 m

Arm		Distal	RIDER_R/L_HME	RIDER_R/L_HLE
R/L Forearm	Cylinder	Proximal	RIDER_R/L_HME	RIDER_R/L_HLE
		Distal	RIDER_R/L_RSP	RIDER_R_USP
R/L Hand	Shpere	Proximal	RWJC_STATIC/LWJC_STATIC	Radius: 0.5*Distance(Rider_R/L_USP,Rider_R/L_RSP)
		Distal	RIDER_R/L_HM2	Radius: 0.5*Distance(Rider_R/L_USP,Rider_R/L_RSP)

Table 12. Equine body segment geometry and defining points to generate horse model. Right and left limbs were labelled as "R" and "L" abbreviations.

Name	Shape	Location	Defining Point 1	Defining Point 2
Head	Cone	Proximal	CRISTA_JOINT_CENTER	Radius: 0.5*Distance(HORSE_FL_CRISTA,HORSE_FR_CRISTA)
		Distal	ATLAS_JOINT_CENTER	Radius: 0.5*Distance(HORSE_L_ATLAS,HORSE_R_ATLAS)
		Extra Target	Anterior, HORSE_T6	-
Thorax	Cone	Proximal	SCAPULA_JOINT_CENTER	ELBOW_JOINT_CENTER
		Distal	HORSE_L3	Radius: 0.5*Distance(HORSE_L_SCAPULA,HORSE_R_SCAPULA)
Lumbar Spine	Cylinder	Proximal	HORSE_S3	Radius: o0.1 m
		Distal	HORSE_L3	Radius: 0.1 m
		Extra Target	Posterior, HORSE_L5	-
Tail	Cylinder	Proximal	HORSE_S3	Radius: 0.5 m
		Distal	HORSE_S5	Radius: 0.05 m
		Extra Target	Anterior, HORSE_L5	-
R/L Upper Arm	Cylinder	Proximal	HORSE_R/L_ELBOW	Radius: 0.075 m
		Distal	HORSE_R/L_CARPUS	Radius: 0.075 m
		Extra Target	Posterior, HORSE_RF/LF_FETLOCK	-
R/L Lower Arm	Cylinder	Proximal	HORSE_R/L_CARPUS	Radius: 0.075 m
		Distal	HORSE_RF/LF_FETLOCK	Radius: 0.075 m
		Extra Target	Posterior, HORSE_RF/LF_HOOF	-
R/L Wrist	Cylinder	Proximal	HORSE_RF/LF_FETLOCK	Radius: 0.075 m
		Distal	HORSE_RF/LF_HOOF	Radius: 0.075 m
		Extra Target	Anterior, HORSE_R/L_CARPUS	-
R/L Thigh	Cylinder	Proximal	HORSE_R/L_STIFLE	Radius: 0.075 m
		Distal	HORSE_R/L_TARSUS	Radius: 0.075 m
		Extra Target	Posterior, HORSE_RH/LH_FETLOCK	-
R/L Lower Leg	Cylinder	Proximal	HORSE_R/L_TARSUS	Radius: 0.075 m
		Distal	HORSE_RH/LH_FETLOCK	Radius: 0.075 m
		Extra Target	Posterior, HORSE_RH/LH_HOOF	-
Pelvis	Cone	Proximal	HORSE_L5	Radius: 0.05 m
		Distal	HORSE_L_HIPE	HORSE_R_HIP
R/L Humerus	Cylinder	Proximal	HORSE_R/L_SHOULDER	Radius: 0.075 mR
		Distal	HORSE_R/L_ELBOW	Radius: 0.075 m
		Extra Target	Posterior, HORSE_R/L_RIB	-
R/L Hoof	Cone	Proximal	HORSE_RH/LH_FETLOCK	Radius: 0.075 m
		Distal	HORSE_RH/LH_HOOF	Radius: 0.075 m

		Extra Target	Posterior, HORSE_R/L_TARUS	-
R/L Femur	Cylinder	Proximal	HORSE_R/L_HIP	Radius: 0.075 m
		Distal	HORSE_R/L_STIFLE	Radius: 0.075 m
		Extra Target	Posterior, HORSE_R/L_TARSUS	-

Following IRB approval, one female participant (21 yo) was recruited from a participant call that was sent to Clemson University Equine Center members and employees. Exclusion criteria included: i) previous history of neurological diseases, ii) any orthopedic or rheumatic condition that could interfere with data quality, iii) known vestibular dysfunction, iv) non-correctable visual disabilities, and v) inhibited trunk or lower limb mobility. Inclusion criteria included i) over 100 hours of horseback riding experience in the past 10 years.

Set up

The three-dimensional motion tracking system (Qualisys, Goteborg, Sweden) will track real time kinematic data. Reflective markers will be placed on human participants at predetermined anatomical landmarks for the horse and human movement trials. Specifically, the software will assess three-dimensional movements across the participant and horses' lumbar spine and pelvis. This provides insight to the pelvis kinematics during equine gait and horseback riding. Specifically, the moment induced on the rider by the horses' lumbosacral joint flexion/extension (Figure 24) will be analyzed to assess the magnitude effect of these movements on rider pelvis kinematics and joint moments. During horseback riding and the reflective markers will track the following:

Table 13. Rider and horse kinematic and kinetic variables that were analyzed from the motion capture and EMG data during data collection

Parameter	Symbol	Method	Method Units
Rider pelvic pitch	(α)	Visual3D Analysis	Degrees (°)
Rider right hip flexion/extension	(γ_R)	Visual3D Analysis	Degrees (°)
Rider left hip flexion/extension	(γ_L)	Visual3D Analysis	Degrees (°)
Rider right hip joint moment	($M_{X/Y/Z,R}$)	Visual3D Analysis	Degrees (°)
Rider left hip joint moment	($M_{X/Y/Z,L}$)	Visual3D Analysis	Degrees (°)
Horse lumbosacral joint flexion/extension	(σ)	Visual3D Analysis	Degrees (°)
Horse lumbosacral joint rotation	(β)	Visual3D Analysis	Degrees (°)
Horse thoracic-lumbar spine length	(L)	Manual Measurement	Meters (m)
Rider muscular activation		Delsys &Visual3D Analysis	Millivolts (mV)

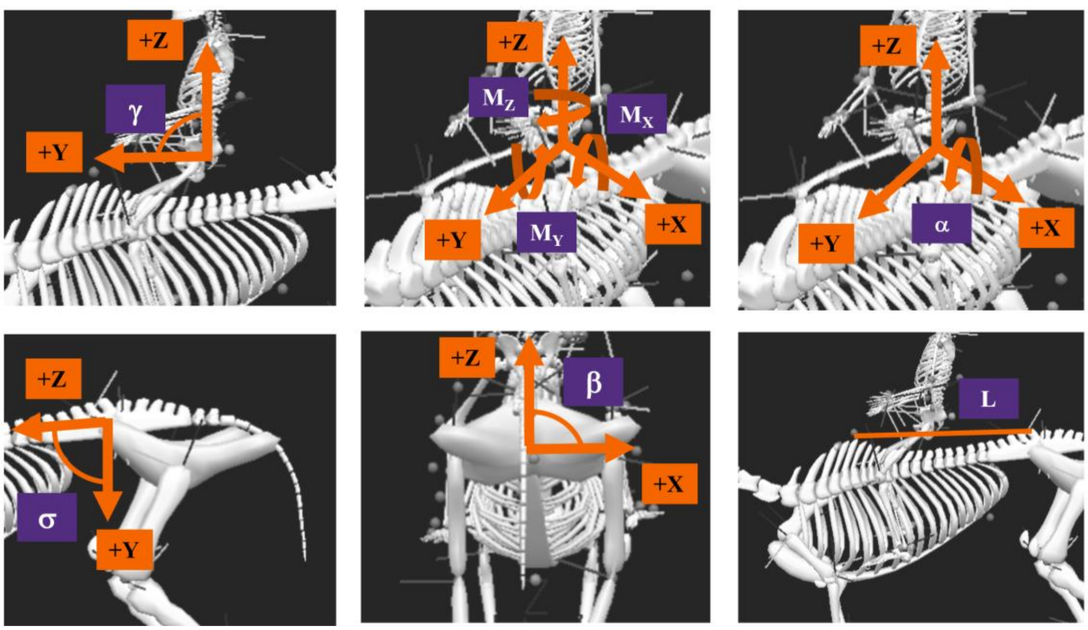

Figure 24. Rider and horse kinematic and kinetic variables in respect to their proper axis and direction of rotation

Wireless EMG (Delsys) sensors were attached to the subjects and provided a real-time reading of the subject's muscle activation patterns during horseback riding. They were placed on their trunk and legs to track muscular contraction in the right and left (1) rectus femoris, (2) iliopsoas, (3) thoracolumbar fascia (4) rectus abdominis, and (5) biceps femoris. These muscle groups were chosen in consistency with previous studies. The sensors were be placed according to SENIAM guidelines[31] and prior studies[32]. All sensors were placed on clean skin with adhesives at desired locations. EMG activity was simultaneously tracked with kinematic motion tracking.

Figure 25. Data collection in outdoor arena with eight motion capture cameras (left) and reflective markers at desired anatomical locations (middle). Horse and rider passed through the motion capture volume to collect kinematic and kinetic data.

Data Processing

All data was exported from Qualisys to Visual3D (C-Motion Research Biomechanics, MD) for analysis. The motion capture data was normalized (0-100% of stride) and filtered for any outliers in the data using an Average and Lowpass Butterworth filter. Desired kinematic data and hip joint moments were calculated and exported to a report through a custom script. Linear cross-correlation was assessed on joint range of motion of the horse's pelvis (σ,β) and rider's pelvic tilt, hip joint flexion/extension, and hip joint moments.

RESULTS

Once data was post-processed and analyzed, general waveform results were generated for desired horse and rider values. In addition to the desired measurements for analysis, horses' pelvis flexion/extension and pelvic tilt was also processed. A single step for a hind leg of the horse requires one phase of flexion and one phase of extension, resulting in a two-beat pelvic tilt pattern (Figure 26). More, the horses' lumbosacral joint movement patterns were also visualized for their rotations about the X and Y axes, flexion/extension and rotation respectively. The horses' thorax experiences two cycles of flexion and extension each and one cycle of positive and one cycle of negative spinal rotation (Figure 27).

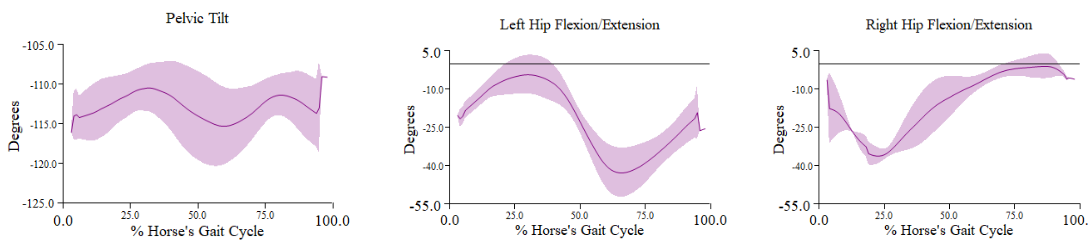

Figure 26. Horse kinematic waveforms to represent pelvis and hip joint patterns during the gait stride. Pelvic tilt of the horse rotates anterior (+) and posterior (-) for two cycles. Hip joints flex (-) and extend (+) for one cycle each.

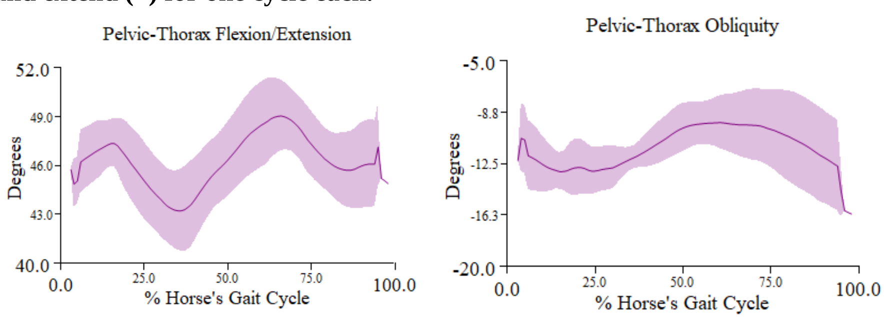

Figure 27. Lumbosacral joint kinematic waveforms. The lumbosacral joint flexes (-) and extends (+) for two cycles during the horse's stride. The lumbosacral joint rotates once positively and once negatively during the gait cycle.

More, similar movement waveforms were generated for rider. The pelvis experienced two cycles of posterior (positive) rotation and two cycles of anterior (negative) rotation. The rider also experiences two cycles of flexion and extension in their hip joint due to the four-beat gait cycle of the horse (Figure 28).

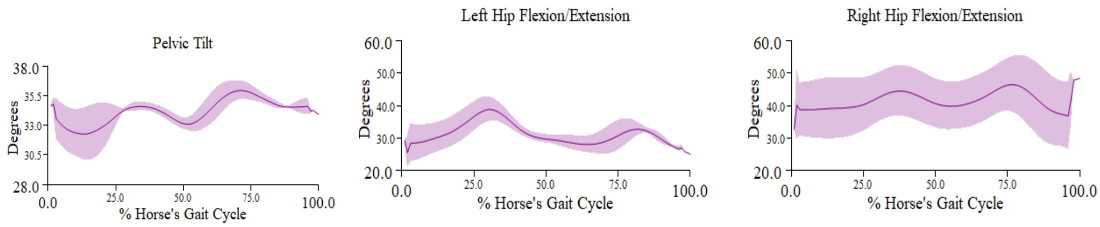

Figure 28. Rider kinematic waveforms to represent pelvis and hip joint patterns during the gait stride. Pelvic tilt of the rider rotates anterior (+) and posterior (-) for two cycles. Hip joints flex (-) and extend (+) for two cycles each.

In addition to the visual representation, peak ranges of motion were calculated from the desired angles and tabulated with the length of each horse's thoracic and lumbar spine (Table 14). The movement induced by the horses' gait to the lumbosacral joint in flexion/extension (σ) and rotation (β) were cross correlated with the responsive movements of the rider's pelvis. These correlation (r^2) values were calculated to interpret the level of influence of the various pelvic-thorax movements of the horse on the rider (Table 15).

Table 14. Kinematic and kinetic variable results following data processing. Values are total ranges of motion expressed by rider and horse joints during 0-100% of the horse's gait cycle.

	α	γ_L	γ_R	$M_{X,L}$	$M_{Y,L}$	$M_{Z,L}$	$M_{X,R}$	$M_{Y,R}$	$M_{Z,R}$	σ	β	L
	Degrees	Degrees	Degrees	N*m/kg	N*m/kg	N*m/kg	N*m/kg	N*m/kg	N*m/kg	Degrees	Degrees	m
Zues	2.186	20.084	22.505	0.223	0.661	0.025	0.284	0.430	0.064	9.200	4.462	0.927
Mallorie	8.411	20.061	14.613	0.169	0.186	0.037	0.000	0.000	0.000	7.738	10.899	0.914
Sierra	3.681	13.776	15.689	0.173	0.156	0.019	0.242	0.285	0.027	6.509	7.140	0.883
Junior	4.412	9.621	11.030	0.140	0.129	0.015	0.200	0.189	0.034	5.784	9.633	0.876
Brody	10.605	17.777	18.075	0.187	0.140	0.027	0.475	0.868	0.170	0.000	0.000	0.864
Ranger	6.028	19.599	15.900	0.147	0.089	0.039	0.283	0.230	0.025	8.073	6.284	0.800

Table 15. Linear correlation between horse kinematic variables (σ,β) and rider kinematic and kinetic responses. Linear correlation is determined by the coefficient of determination (r^2).

	α	γ_L	γ_R	$M_{X,L}$	$M_{Y,L}$	$M_{Z,L}$	$M_{X,R}$	$M_{Y,R}$	$M_{Z,R}$
σ	0.007	0.837	0.765	0.509	0.497	0.349	0.867	0.608	0.429
β	0.543	0.111	0.655	0.356	0.372	0.001	0.882	0.734	0.360

These correlations are also visible in Figure 29-Figure 36. The flexion/extension of the horse's lumbosacral joint (σ) had a significant, positive relationship with rider's left (r^2=0.837) and right (r^2=0.765) hip joint flexion/extension range of motion ($\gamma_{L/R}$). It also had positive correlation (r^2=0.867) with the rider's right hip joint moment ($M_{X,R}$). Therefore, as the horse's lumbosacral joint flexion/extension increased, there was a conclusive increase in rider hip joint flexion and extension and a possible increase in rider joint moments, though only evident in the right hip. The remaining relationships were not strong enough to draw conclusions. More, the lumbosacral rotation (β) also had a significant negative correlation (r^2=0.882) with the rider's right hip joint moment ($M_{X,R}$ and $M_{Y,R}$). As the lumbosacral joint rotation increased, this resulted in a decreased joint moment experienced by the rider. While no other correlations were significant, there were evident negative trends for all rider pelvis relationships, other than pelvic tilt, when compared to lumbosacral joint rotation.

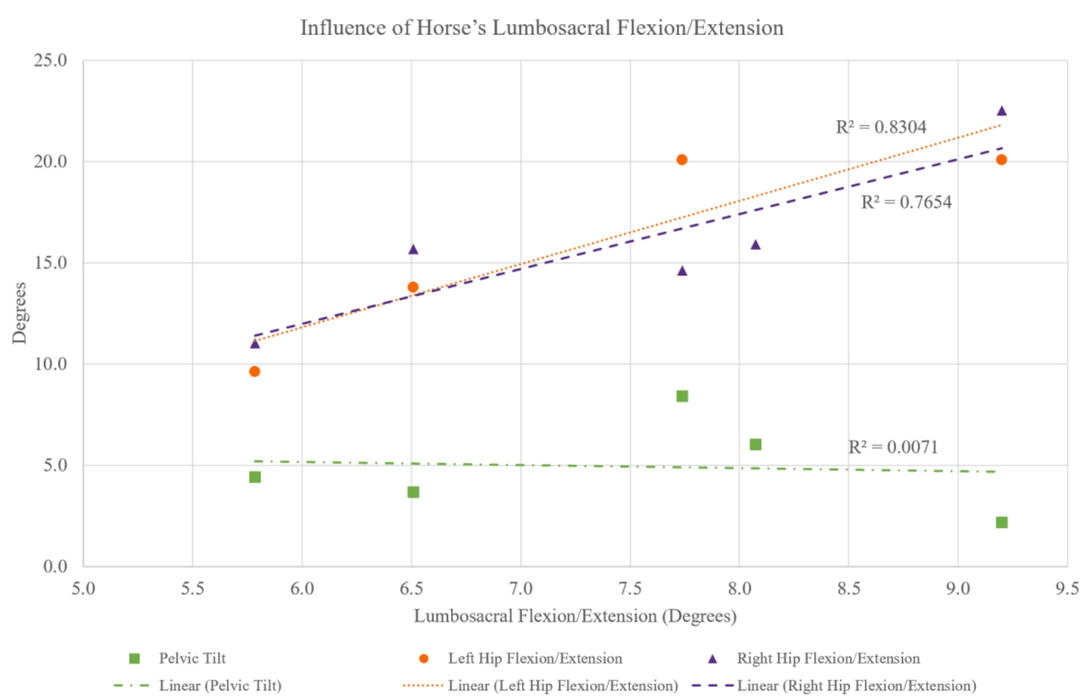

Figure 29. Linear correlation between the horses' lumbosacral flexion and extension and rider's left hip flexion and extension (orange, dot), right hip flexion and extension (purple, dash), and pelvic tilt (green, dot-dash).

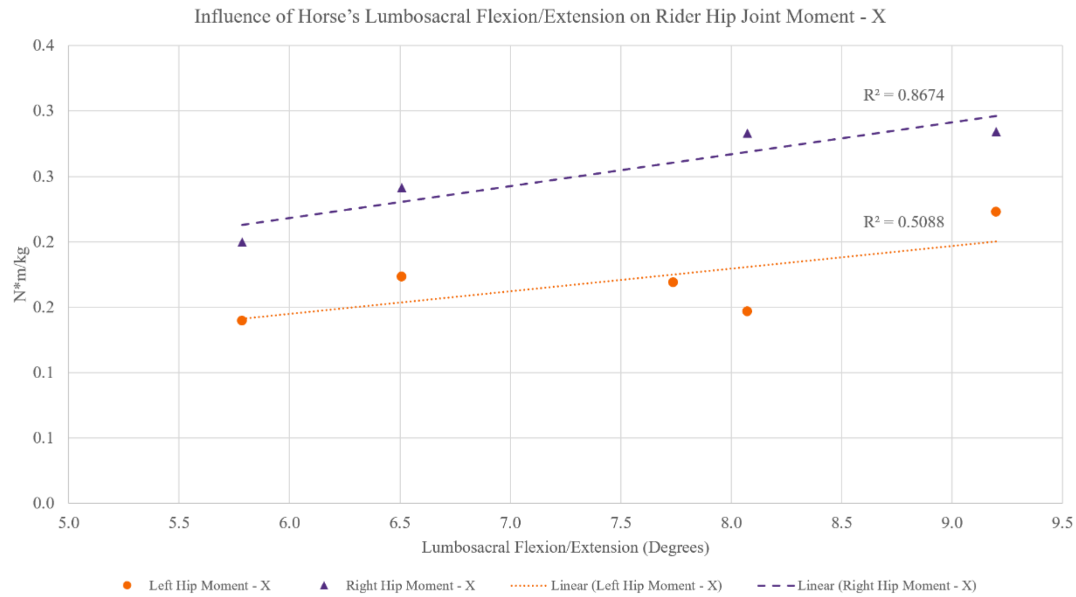

Figure 30. Linear correlation between the horses' lumbosacral flexion and extension and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the x-axis.

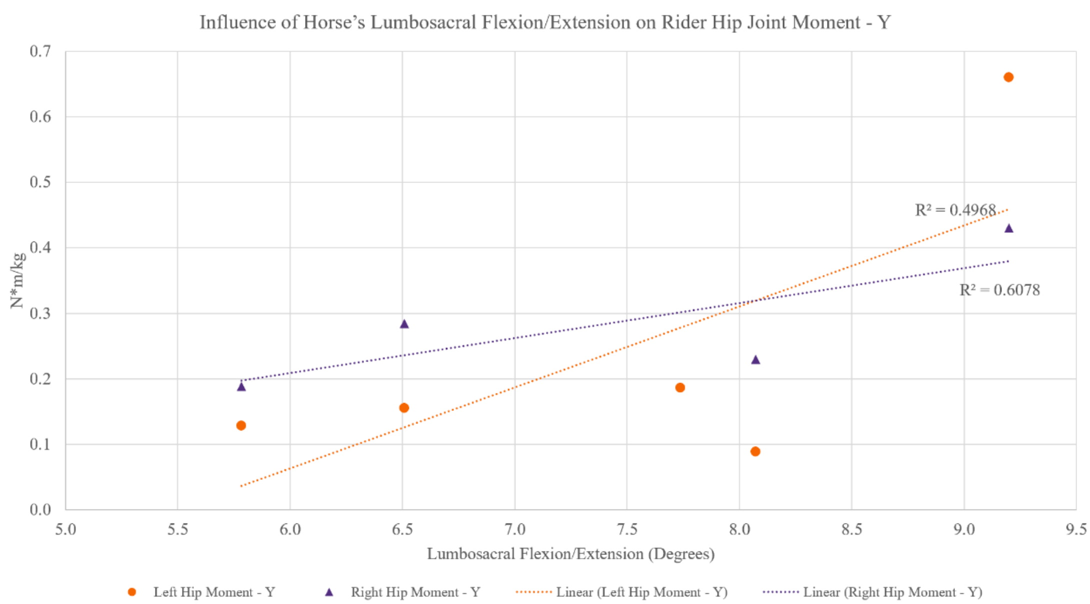

Figure 31. Linear correlation between the horses' lumbosacral flexion and extension and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the y-axis.

Figure 32. Linear correlation between the horses' lumbosacral flexion and extension and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the z-axis.

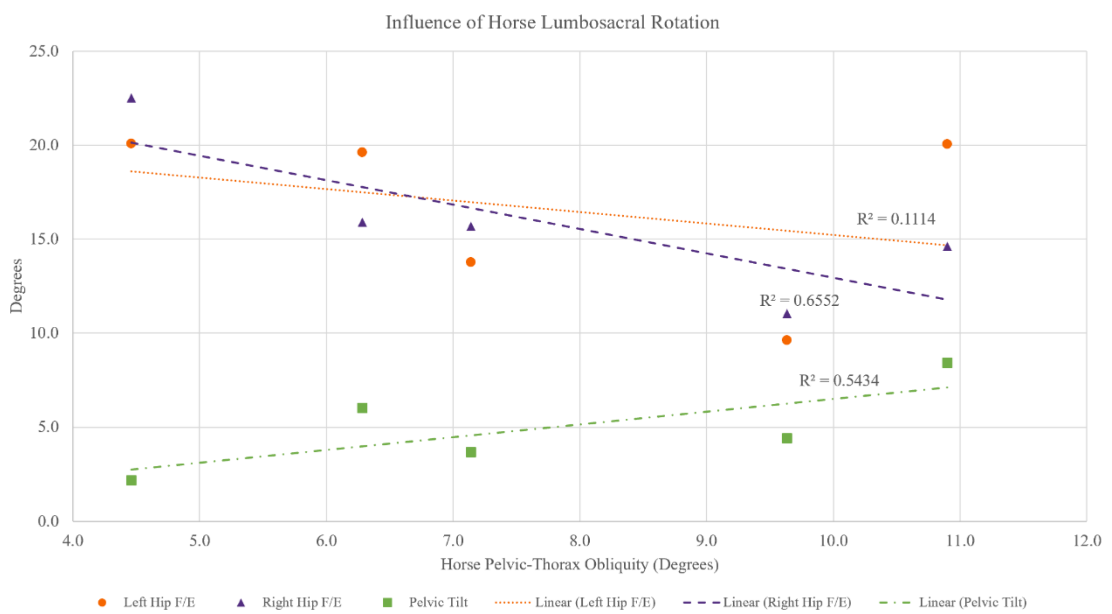

Figure 33. Linear correlation between the horses' lumbosacral rotation and rider's left hip flexion and extension (orange, dot), right hip flexion and extension (purple, dash), and pelvic tilt (green, dot-dash).

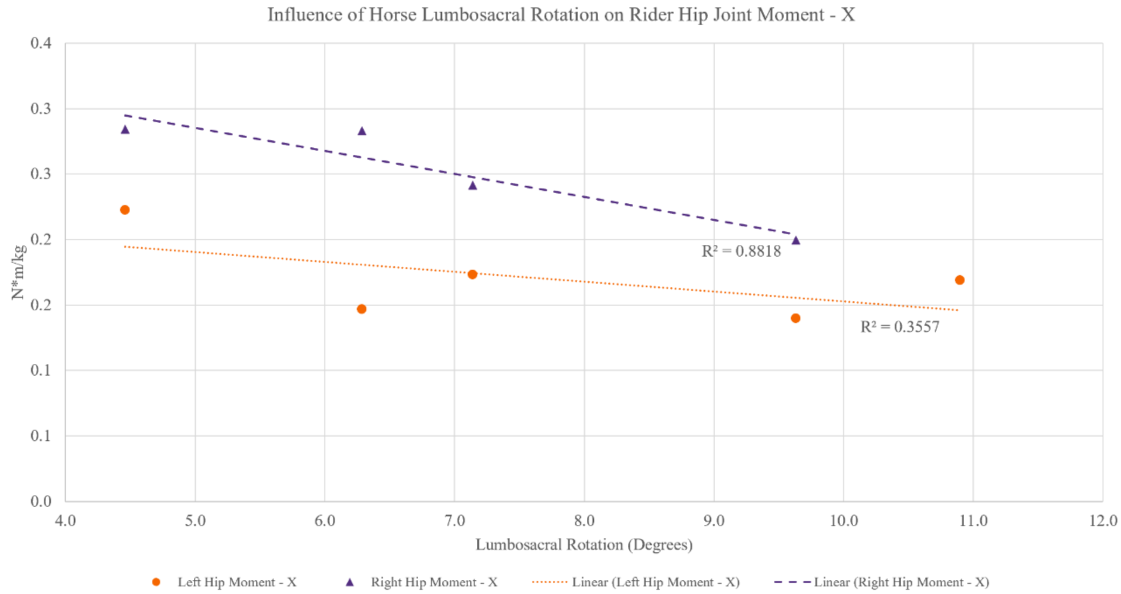

Figure 34. Linear correlation between the horses' lumbosacral rotation and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the x-axis.

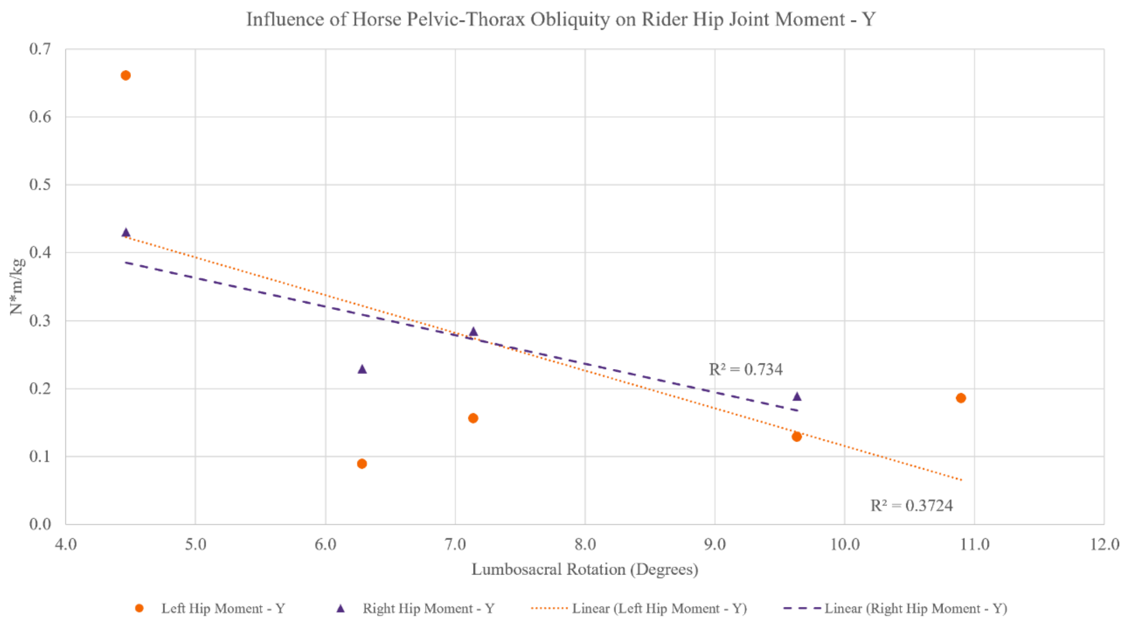

Figure 35. Linear correlation between the horses' lumbosacral rotation and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the y-axis.

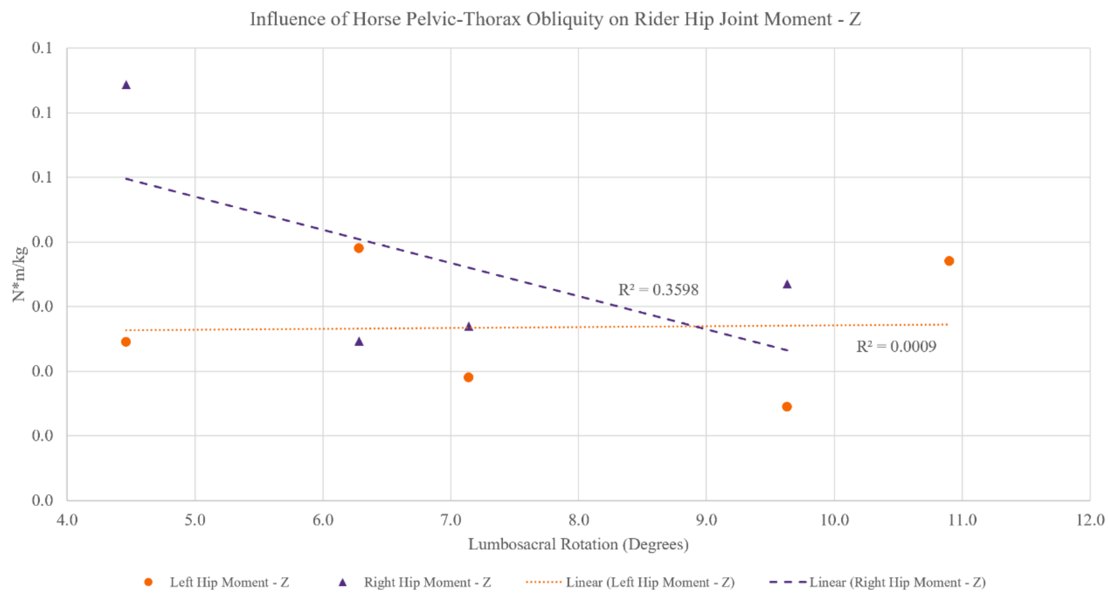

Figure 36. Linear correlation between the horses' lumbosacral rotation and rider's left hip joint moment (orange, dot), right hip joint moment (purple, dash) about the z-axis.

DISCUSSION

Relative movement patterns of the horse's hip flexion/extension were consistent with patterns and magnitudes previously found in literature (Byström, Rhodin, von Peinen, Weishaupt, & Roepstorff, 2010; Eckardt & Witte, 2017b; Goodworth, Barrett, Rylander, & Garner, 2018b; Münz, Eckardt, Heipertz-Hengst, Peham, & Witte, 2013b; Severyn et al., 2022; Wang et al., 2018b). Spinal flexion/extension and rotation of the lumbosacral joint also presented similar movement patterns (Audigie, Pourcelot, Degueurce, Denoix, & Geiger, 1999; Byström et al., 2010; Clayton & Hobbs, 2017; Haussler, Bertram, Gellman, & Hermanson, 2001; Tabsuri, Thawinchai, Peansukmanee, & Lugade, 2021). The lumbosacral junction range is a result of the linkage system of the pelvis through the hind legs, and may be affected by ground reaction forces and step length. Muscularly, this joint is directly associated with the muscle groups in the lumbosacral, pelvic, neck, and lumbar regions depending on the point in the stride (Gallacher et al., 2013). Increased development of these regions could assist in the stabilization of the lumbosacral joint and protect the horse from further back problems (Gallacher et al., 2013).

Magnitudes of lumbosacral range of motion are variable within literature, based on the segment the lumbar spine is referenced against. Results in literature (Audigie et al., 1999; Johnson & Moore-Colyer, 2009) found a similar two-cycle flexion and extension of this joint, but a smaller range of motion than those found in this study. However, they measured the lumbosacral joint in relation to the sacral spine, not the pelvis. Similar to this study, when measured relative to the pelvis, the lumbosacral joint expresses slightly larger ranges of motion in flexion and extension (Tabsuri et al., 2021). This is likely due to the immediate impact the linkage system, from the hind hoof to the horse's hip joint, has on the junction between the pelvis and the lumbar spine. This angular reference represents a more accurate depiction of the effects of joint reactions from ground forces and step length, because of its continuous linkage system.

The rider's 3-dimensional pelvic rotations are believed to be directly linked, though with a slight phase delay, to the 3-dimensional rotations of the horse's thorax (Byström et al., 2010; Clayton & Hobbs, 2017; Münz et al., 2014a). However, magnitudes of these movement patterns and their corresponding response by the rider have not been largely explored. Due to the flexion/extension of the lumbosacral joint, the thorax of the horse directly integrates with the rider's pelvis generating a similar responsory movement.

This study determined there was a strong correlation between the magnitude of range of motion for the lumbosacral joint (σ) and the magnitude of the hip joint range in the rider ($\gamma_{R/L}$). This range references the pelvis as it rotates anterior and posterior in relation to the corresponding thigh of the rider, therefore this flexion and extension incorporates the relative movements of the thighs during the horses' walk as well. Therefore, the larger the flexion/extension of the horse's lumbosacral joint, there's a strong likelihood the rider will express greater flexion and extension in their hip joints. More, this rotation of the lumbosacral joint expressed a positive relationship between the magnitude of the joint range and the rotational moment experienced in the rider's hip joint. There was only a significant relationship between the horse joint (σ) and the rider's moment about the x-axis ($M_{X,R}$) in the right hip joint. However, due to generally positive trends, it is believed that a larger data set would result in a stronger correlation between the lumbosacral joint flexion/extension and the rotational moments experienced by the rider, especially about the x-axis ($M_{X,R}$) and y-axis ($M_{Y,R}$).

Additionally, the rotational magnitude of the lumbosacral joint (β) also exhibited a strong correlation with the joint moment about the x-axis ($M_{X,R}$) and y-axis ($M_{Y,R}$). These trends showed a decrease in rider joint responses as the lumbosacral joint rotation increased across various horses. As the horse's hind limb lifts, the horse's trunk and rider's pelvis will rotate (roll) away from the limb in stance (Byström et al., 2010; Clayton & Hobbs, 2017; Münz et al., 2014a). This movement is similar to pelvic obliquity when referring to human gait. As this rotation increases in the horse, it would be expected to see an increase flexion/extension on the rider hip joint ipsilateral to the hind leg in stance, due to the larger pelvic rotation (roll) toward the hind leg as it swings. However, also happening simultaneously, the forelimb (contralateral to the hind limb in its swing phase) contacts the ground and the rider's pelvis rotates anteriorly resulting in flexion of the hip joints.

More, there was a modest positive relationship between lumbosacral rotation and the pelvic tilt of the rider (r^2=0.543). Therefore, the lumbosacral rotation of the horse may not cause a decrease in range of motion of the rider's pelvis anteriorly/posteriorly. Rather, the decreasing relationship was only seen among the cross correlation with hip joint flexion/extension and hip joint moments. This is significant because these joint analyses are all based on the relationship between the pelvis and the rider's femur. Therefore, the increase in lumbosacral rotation of the horse's pelvis, eliciting a similar rotation of the thorax (Clayton & Hobbs, 2017) could cause the rider to utilize their thighs to maintain postural control and, ultimately, decrease the measured movements of the hip joints.

This could have implications when choosing a horse for use in equine assisted activities and therapies (EAAT) or hippotherapy (HPOT) because horses with great lumbosacral joint rotations may induce muscular stimulation of the legs in the rider. This may be beneficial for specific populations that desire to focus on lower limb strength. More, decreased joint moments also signify less muscular activation due to a decrease in force acting on the joint. Again, this could be desirable if the focus of the horseback riding intervention is on the lower limbs. In contrast, an expert may choose a horse with a smaller lumbosacral joint rotation in order to maximize the flexion/extension and joint moments of the rider's hips in order to focus on activation of the trunk muscles for specific population groups. However, it should be emphasized that this rider was healthy and experienced, therefore had above average control while horseback.

CONCLUSION

Overall, the model was successful in tracking rider and horse kinematics, and rider joint moments. There were a few strong correlations found between the lumbosacral joint of the horse and the resulting movements in the rider, including hip flexion/extension and some joint moments. While only a few significant correlations were measured, these relationships followed consistent patterns and further investigation with a larger sample size could expand on these correlations. There may be a connection between increased lumbosacral joint rotation (roll) and decreased hip joint flexion/extension and joint moments due to increased tension in the rider's thighs to maintain postural control while horseback. However, this needs to be investigated further. Future work with EMG signals could also provide necessary insight to this work and will be further incorporated to future studies.

CHAPTER SIX

NOVEL 3D MODEL GENERATION OF WHEELCHAIR TENNIS ATHLETE AND POWER EFFICIENCY ANALYSIS OF TWO DRILL TECHNIQUES

INTRODUCTION

Current mobility assessments in adaptive sports

Adaptive activities are largely under researched in the field of biomechanics, especially in adaptive sports. Adaptive sports are becoming increasingly popular around the world in participation and interest. The 2016 Rio Paralympic Games were viewed by 4.1 billion people, a 127% increase from the 2004 Athens Paralympic Games (International Paralympic Committee, 2017). Adaptive sports offer many benefits to people living with mobility limitations such as overall increased quality of life and satisfaction (Côté-Leclerc et al., 2017; Yazicioglu, Yavuz, Goktepe, & Tan, 2012), health (Zabriskie, Lundberg, & Groff, 2005), community involvement, and social interaction (Sundar, Brucker, Pollack, & Chang, 2016). Wheelchair tennis is among the most popular adaptive sports in the world for those with functional impairments. Mobility is a widely investigated area regarding wheelchair tennis and is considered “the single most important aspect” (Wayne, 2002) in mastering the sport. While there is research on the performance of wheelchair tennis players utilizing inertial measurement units (Rietveld et al., 2019; Van Der Slikke, Berger, Bregman, & Veeger, 2015), the mobility and efficiency of specific routes taken during a match have yet to be explored using 3D video motion capture. Several of these studies have explored topics such as propulsion (Vanlandewijck, Theisen, & Daly, 2001), skid correction (Van Der Slikke et al., 2015), and chair velocity with and without a racquet (Goosey-Tolfrey & Moss, 2005). Due to the scarce 3D motion capture data collected within this sport, there is a need to create a singular player model encompassing the player and wheelchair as a unit. The creation of this model will enable kinematic analysis athletes who are already at a biomechanical disadvantage.

Sports performance research and 3D modelling

3D modeling has gained traction in the sports performance industry, and has been used by a countless number of sports programs to analyze physical condition, athletic performance, injury mechanism, and prevention and rehabilitation for an athlete (Pueo & Jimenez-Olmedo, 2017). For example, a study was conducted on competitive male runners using 3D modeling to examine the ankle and knee kinetics between strike patterns in order to determine which strike pattern is the premier foot strike pattern and which patterns are at more risk of injury (Kuhman, Melcher, & Paquette, 2015). The use of 3D motion capture is becoming more prevalent among major sports leagues, with multiple MLB teams utilizing it to improve performance of pitchers and to prevent injury (Kuhman et al., 2015). There are three different methods for 3D motion capture: optical system, non-optical system, and marker-less system. Optical systems use reflective spheres that are placed on major anatomical landmarks of the athlete and, in the case of this study, the wheelchair. Motion capture cameras were used to capture the athlete-wheelchair marker set. Non-optical systems use inertial measurement unit sensors attached to landmarks on the body to track their position. Marker-less systems are becoming more popular, as they use sophisticated computer image processing to create the 3D model without the use of reflective spheres (Pueo & Jimenez-Olmedo, 2017). The purpose of this study was to examine the mobility and efficiency of wheelchair tennis athletes between two drill assessments using 3D motion capture technology with the intent to produce the first 3D wheelchair athlete model, including the racquet and the specifications to create it.

RESEARCH METHODS

Generation of wheelchair tennis athlete model

A novel wheelchair tennis athlete biomechanical model was created using 3D video motion capture technologies (Qualisys) and Visual3D. For the human skeletal model, the pelvis was excluded due to hindered visibility from the wheelchair and legs to be easily adaptable to all athletes. Furthermore, the models were built with consideration for kinetic and kinematic outcomes including angular velocity of the racquet head, shoulder joint rotation, and the kinetic energy of the athlete-wheelchair system. For the unprecedented wheelchair, markers were placed on common shafts, the seat, and the wheels. Similarly, markers were placed on the proximal and distal end of the racquet and along the shaft (Figure 37). Initial data was collected in the lab to create 3D models in Qualisys and Visual3D.

Within Visual3D, the head and thorax portions of the athlete follow the Visual 3D RAB Upper Extremity Model (Rab et al., 2002). Segments were modelled as cylinders and the head as an ellipsoid (Figure 37). The racquet was modelled as a cone (Figure 37). Further, the wheelchair shafts were modelled as cylinders, the wheels as ellipsoids, and the seat as a cone to compensate for the lower body weight and center of gravity of the entire system (Figure 37). Object files of the head, thorax, arms, wheelchair, and racquet were overlayed and scaled to create a visual representation with no effect on the biomechanical calculations. This model creation was executed on preliminary data. Final data collection on the study's athletes was executed after the model was created and validated.

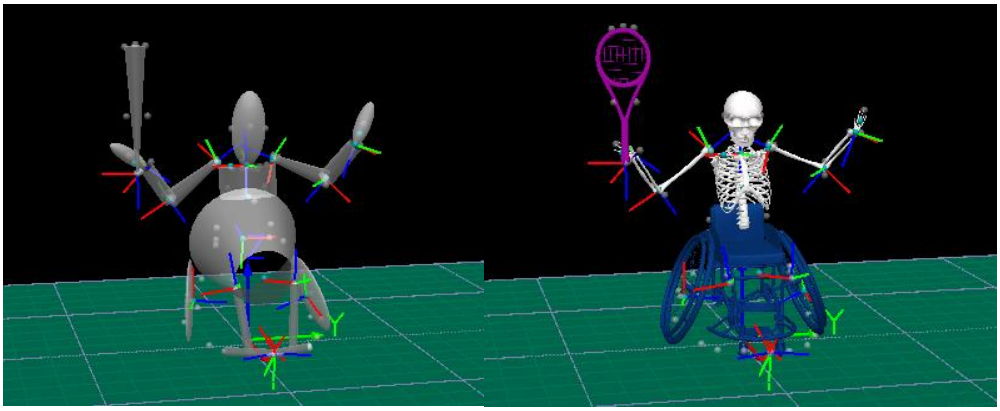

Figure 37. Geometric creation and orientation of the wheelchair athlete system (left) and its graphical overlay to represent real-life visuals (right).

Table 16. Model generation of wheelchair tennis athlete upper body including representative geometric shapes and defining anatomical locations

Name	Shape	Location	Defining Point 1	Defining Point 2
Racquet	Cone	Proximal	Medial: RAC_BOTTOM2	Lateral: RAC_BOTTOM1
		Distal	Medial: RAC_TOP2	Lateral: RAC_TOP1

Table 17. Model generation of tennis racquet including its representative shape and defining locations

Name	Shape	Location	Defining Point 1	Defining Point 2
R/L Upper Arm	Cylinder	Proximal	Center: (R/L)SHLDR2	Radius: 0.0214 m
		Distal	Lateral: (R/L)LH	Medial: (R/L)MH
R/L Forearm	Cylinder	Proximal	Lateral: (R/L)LH	Medial: (R/L)MH
		Distal	Lateral: (R/L)DR	Medial: (R/L)DU
R/L Hand	Cylinder	Proximal	Lateral: (R/L)DR	Medial: (R/L)DU
		Distal	Joint Center: (R/L)MC3	Radius: 0.5*distance((R/L)DR,(R/L)DU)
Thorax	Cylinder	Proximal	Center: SC_C7	Radius: 0.5*distance(RAC,LAC)
		Distal	Center: XP_T8	Radius: 0.5*distance(RAC,LAC)
Head	Ellipsoid	Proximal	Center: C7_STERN	Radius: 0.01 m

		Distal	Lateral: R_EAR	Medial: L_EAR

Table 18. Model generation of the sport wheelchair including its representative shapes and defining locations

Name	Shape	Location	Defining Point 1	Defining Point 2
R/L Lateral Wheel	Ellipsoid	Proximal	Lateral: (R/L)W_PRONG1	Medial: (R/L)W_PRONG2
		Distal	Center: (R/L)W_CENTER	Radius: 0.2461 m
Seat	Cone	Proximal	Center: WC_seatbackcenter	Radius: 0.5*distance(L_SEATBOTTOM, R_SEATBOTTOM)
		Distal	Lateral: L_FRAME1	Medial: R_FRAME1
R/L Vertical Shaft	Cylinder	Proximal	Center: (R/L)_FRAME2	Radius: 0.0254 m
		Distal	Center: (R/L)_FRAME3	Radius: 0.0254m
		Extra Target	Medial, (R/L)_FRAMECENTER	-
R/L Horizontal Shaft	Cylinder	Proximal	Center: (R/L)_FRAMECENTER	Radius: 0.024 m
		Distal	Center: (R/L)_FRAME4	Radius: 0.024 m
		Extra Target	Lateral, (R/L)_FRAME4	-

Study design and patient recruitment

Four wheelchair tennis athletes executed a novel version of a common match-simulation drill (the Figure 8 drill), which they called the Pyramid drill. This version of the drill focused on attacking the net rather than staying in the backcourt while volleying with their teammate or coach. For the purposes of the study, we wanted to capture three common maneuvers in the drill: a forehand, backhand, and short ball swing. The outline of drill simulations is seen in Figure 38**.** Eight motion capture cameras were placed strategically around a tennis court to capture the volume of the area of movement (one side of the tennis court only).

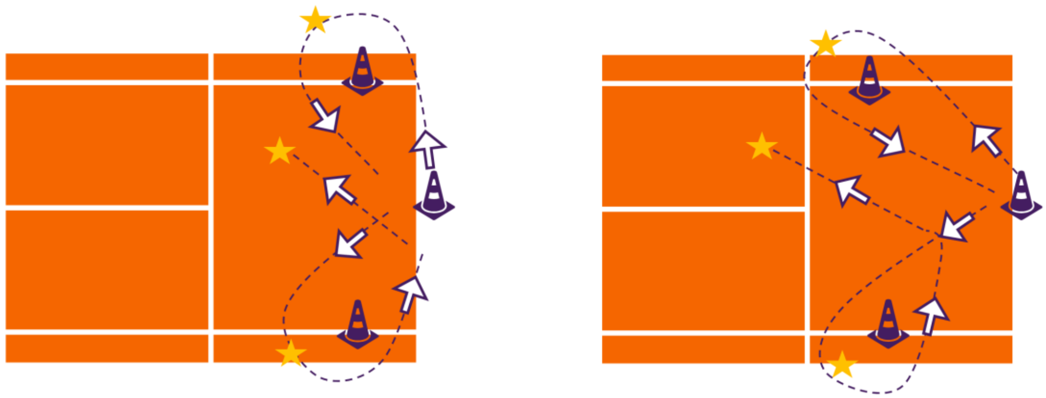

Figure 38. Graphical representation of the Figure 8 (left) and Pyramid (right) drills. Each star represents a hit: forehand, backhand, and short ball.

Figure 39. Data collection of wheelchair tennis athletes with reflective markers at desired anatomical locations on the human and segment locations of the racquet and wheelchair.

Data processing

After 3D motion capture of the two drills (Pyramid and Figure 8), the data captured was gap filled in Qualisys. An Automatic Identification of Markers (AIM) model was applied, and each marker was applied to its correct designation to allow the AIM model to create segments between markers to create the athlete, wheelchair, and racquet segmented model in Qualisys. The gap filled Qualisys model was then exported to Visual 3D where kinematic calculations took place. Once exported, tags were created for the Figure 8 Hub drill and the Pyramid drill and the corresponding data files were placed under the appropriate tag name. A static file for each athlete was exported into Visual 3D. The 3D model was then created using the static file, and then that model was assigned the Figure 8 and Pyramid motion files. Events were added to each motion file after the model was assigned. Events were identified at the start of the forehand, backhand, and short ball swing and at the conclusion of each swing. These events break up the motion file allowing the kinematics of each swing to be analyzed. A pipeline was then used to calculate important kinetic and kinematic outcomes including angular velocity of the racquet head, shoulder joint rotation, and the kinetic energy of the athlete-wheelchair system at each swing define by the events created. The pipeline was executed, and data was exported into a text file. That text file was then entered into Excel where the data was manipulated to compare the kinetics and kinematics of the Figure 8 drill and the Pyramid drill at each swing. Total energy of the Figure 8 and Pyramid drill was calculated to compare the differences. This was calculated in Visual 3D by creating only two events, one at the beginning of the trial and one at the end of the short ball swing. The same pipeline was used except calculations not solving for energy were omitted. These results were entered into Excel where graphs were created to compare the total energy of the two drills.

RESULTS

Racquet velocity and shoulder joint kinematics

Once the model was validated, Athlete 1, 3, 4, and 5 data was collected and processed. Each athlete's movement profile was generated which included shoulder joint angle (in reference to the thorax) and racquet velocity (in reference to the thorax), both in the X, Y, and Z directions for the forehand, backhand, and short ball. Data was visually compared between average kinematic patterns of the Pyramid and Figure 8 drills. A sample of the data reports can be seen in Figure 40. More, peak absolute velocities were calculated for each swing in the Figure 8 (Table 19) and Pyramid (Table 20).

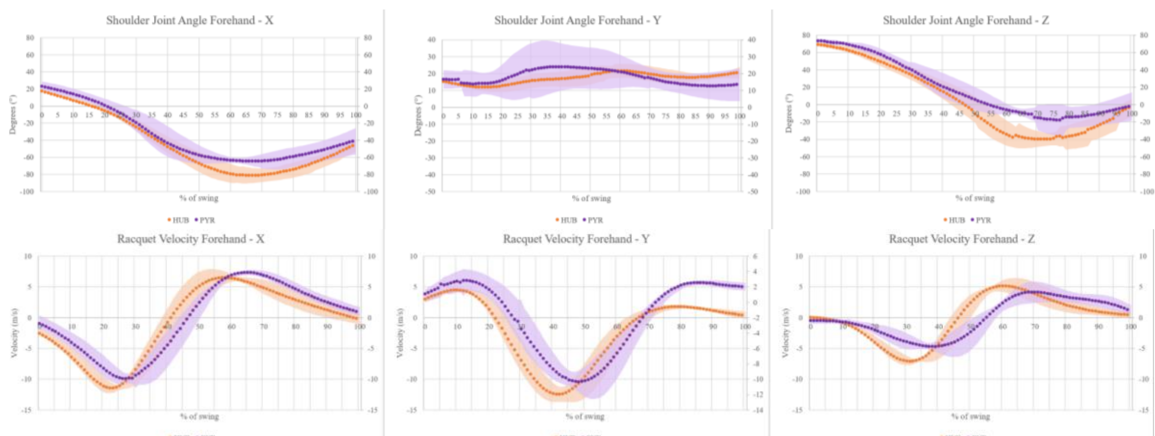

Figure 40. Sample set of the athlete data reports generated by the athlete's data profiles after pipeline execution and analysis. Data represented is Shoulder Joint Angle (°) in the first row and Racquet Velocity (m/s) in the second row, for the forehand swing of the Figure 8 (orange) and Pyramid (purple) drills in the X, Y, and Z axes. Not visually represented, but included in the report, are the same analysis for the backhand and short ball swings. Equivalent reports were generated for Athletes 1, 3, and 5.

Table 19. Figure 8 maximum racquet velocities for all directions (X, Y, Z) for all athletes

Hit	Variable	Direction	Athlete 1	Athlete 3	Athlete 4	Athlete 5
Forehand	Velocity	X	9.70±2.49	9.90±2.61	11.46±0.74	9.40±8.77
		Y	7.40±5.01	6.70±2.31	12.47±1.22	11.22±6.48
		Z	5.93±0.97	3.30±0.92	7.12±0.45	4.81±2.85
Backhand	Velocity	X	10.59±4.23	11.26±1.01	10.88±1.35	10.84±3.50
		Y	10.30±2.42	4.74±0.34	9.50±2.42	11.71±1.37
		Z	5.87±0.81	4.49±2.27	3.95±2.21	5.27±1.78
Short Ball	Velocity	X	11.00±3.86	8.83±3.19	10.40±2.02	9.97±4.73

		Y	10.92±4.89	4.55±1.69	10.66±3.07	9.59±5.49
		Z	6.65±2.49	1.82±2.56	7.09±2.43	5.72±2.39

Table 20. Pyramid maximum racquet velocities for all directions (X, Y, Z) for all athletes

Hit	Variable	Direction	Athlete 1	Athlete 3	Athlete 4	Athlete 5
Forehand	Velocity (m/s)	X	8.79±2.63	6.76±2.96	9.91±0.81	11.14±8.28
		Y	7.26±5.09	5.30±2.16	10.34±1.70	8.75±3.49
		Z	10.42±4.79	0.83±0.33	4.70±0.84	6.38±5.24
Backhand	Velocity (m/s)	X	11.78±6.02	7.08±3.98	9.08±0.81	13.04±3.15
		Y	7.20±10.06	3.27±1.78	9.10±2.37	11.03±2.54
		Z	6.87±2.80	3.18±2.38	6.00±1.44	3.11±2.06
Short Ball	Velocity (m/s)	X	8.69±3.83	7.71±1.65	10.74±1.04	11.98±1.82
		Y	8.56±3.44	5.06±0.69	11.25±1.64	11.36±1.74
		Z	23.71±35.42	1.58±0.73	6.58±1.54	6.25±1.99

Comparison of total energy conservation

In addition to movement kinematic data, total energy of the wheelchair-athlete system was calculated for the entire drill (start of movement through short ball swing). These trends are visible in Figure 41, where climb of a peak is the gain of kinetic energy coming into the swing, the peak is the kinetic, turning, and rotational energy maximum of the swing, and the decline of the peak is the loss of those energies (likely due to friction and stopping on the turn by the athlete). As expected, flatter peaks are visible in the Figure 8 drill due to the wider turns, while the Pyramid drill has visibly sharper peaks (turns).

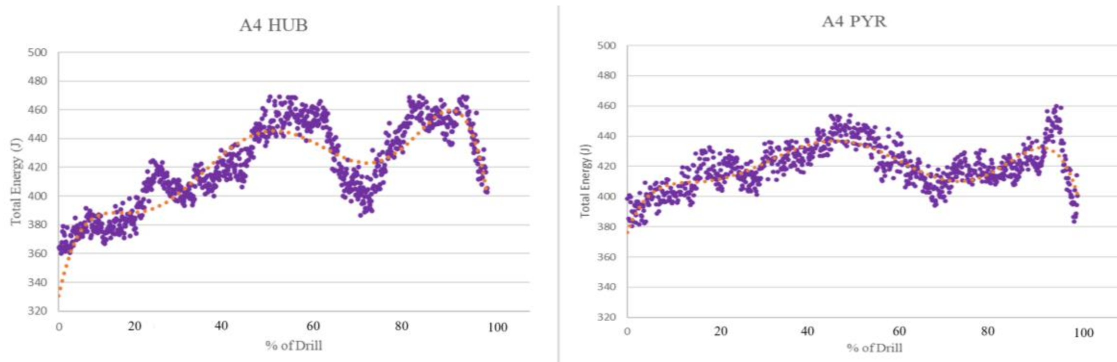

Figure 41. Figure 8 ("HUB", left) Pyramid (right) Total Energy (J) trends with visible peaks occurring during the various swing phases.

Maximum energy peaks were evaluated between the Figure 8 and Pyramid drill, as well as the corresponding range from the maximum range to the minimum value within the drill. The maximum energy represents the total kinetic, rotational, and turning energy of the system. The range of energy numerically signifies the maximum change in energy fluctuation and expenditure experienced by the athlete during the drill. It was found that Athlete 1 had a significantly larger maximum energy peak in the Pyramid drill, and Athlete 1 and Athlete 3 had resultingly larger ranges of energy usage. However, Athlete 4 and Athlete 5 had significantly smaller energy peaks in the Pyramid drill, but significantly smaller energy expenditure ranges.

Table 21. Maximum energy peaks and total range of energy expenditure for all athletes between the Figure 8 and Hub drills. Significantly larger maximum and range values are represented by *.

	Athlete 1		Athlete 3		Athlete 4		Athlete 5	
Drill	**Max (J)**	**Range (J)**	**Max (J)**	**Range (J)**	**Max (J)**	**Range (J)**	**Max (J)**	**Range (J)**
Figure 8	489.20	95.73	409.20	59.02	469.56*	109.37*	428.21*	87.78*
Pyramid	494.44*	111.49*	407.35	81.35*	459.78	79.22	399.18	54.15

DISCUSSION

This study offers the first 3D wheelchair athlete model for use in motion capture analysis, and provided evaluation of mobility and efficiency among athletes performing common tennis movements (i.e., forehand, backhand, and short swing). Differences were seen between athletes in this study. Most notably, the novice players and those that were not familiar with the Pyramid drill did not see appreciable differences in racket velocities between the two drills. However, the main objective was not to assess differences among athletes but internal differences between drills. For racket speeds, differences in phase (when peak speeds were seen) were predominantly seen between the Pyramid and Figure 8 drill. In the Pyramid drills, athletes 4 and 5 were shown to shift their peak ball contact speeds earlier in their swing phase with respect to the Figure 8 drill. This could be hypothesized to occur because the athletes would be anticipated to execute a sharper turn radius in the Pyramid drill than the Figure 8 drill, resulting in a shorter swing phase. Alternatively, Athlete 1 shifted their maximum swing velocities later in their swing phase during the Pyramid Drill as compared to the Figure 8. This might be a result of over-effort or unfamiliarity with the new wheelchair pattern or expected more abrupt changes in direction in the Pyramid Drill. This could be hypothesized to cause less attention, concentration and duration of the swing when the hand is not on the chair for directional control.

Peak velocities found by a study (Landlinger, Lindinger, Stöggl, Wagner, & MüLler, 2010) in match-comparable cross court and down the line scenarios were 32.3±1.9 and 29.8±1.5 m/s, respectively for elite able-bodied athletes and 30.4±0.8 and 27.8±1.3 m/s, respectively for high performance athletes. These values are considerably higher than the values in the current study because these athletes were able-bodied and with elite and high performance rankings. However, if you compare the values from

Landlinger et al (Landlinger et al., 2010) with a similar scale between wheelchair tennis athletes and able-bodied athletes in a study by Reid, Elliott, and Alderson (Reid, Elliott, & Alderson, 2007) there was approximately a 10-13 m/s difference in maximum velocities during tennis serves. Assuming this as a standard difference between the two population groups, and considering the athletes in the current study were novice and amateurs, the maximum velocities calculated from the model in this study were considered to be within a confident range.

Peak energy expenditure for this study all athletes ranged from 399.18 [J] to 494.44 [J]. Another study (Barbosa, Forte, Estrela, & Coelho, 2016) evaluated peak energy cost in a wheelchair sprinter which ranged from 95.67 [J/m] to 100.14 [J/m] depending on the athlete's position. When estimating our maximum values across 5 meters, approximately the distance between cones in the Figure 8 and Pyramid drill, and considering the difference in athlete experience, the maximum values of the athletes in this study are comparable to those found in Barbosa et al. (Barbosa et al., 2016). Energy Cost during the cycling phases in another study (Forte, Marinho, Morais, Morouço, & Barbosa, 2019) ranged from 33.33 [J/m] to 276.26 [J/m], depending on the phase of the cycling pattern. Again, when approximating this study's [J/m], these values validate the findings found with the athletes in this study.

For maximum energy peaks and expenditures, there was an overriding trend between and within drill type. Some athletes had higher energy peaks and expenditures in the Pyramid Drill, and others showed lower values, as compared to the Figure 8. Athletes 1 and 3 had no prior experience with the Pyramid drill, and Athlete 1 exerted significantly larger maximum energy expenditure in the Pyramid drill than the Figure 8. Both Athletes 1 and 3 experienced larger ranges in energy usage in the Pyramid as well. However, Athletes 4 and 5, who both practiced the Pyramid drill regularly, exerted significantly larger maximum energies and ranges in the Figure 8. It could be hypothesized that this is a result of experience level, fitness, or familiarity, but it is unclear from this work. In addition, the smaller ranges in energy usage could represent greater energy conservation during turns. This would be impactful for athletes practicing this form in order to conserve energy during drills, and ultimately, real-time matches. More work is needed to clarify this finding. It was seen, however, that the maximum peak energy exerted by the athlete always coincided with the largest range of energy expenditure. This is a likely correlation because larger peak values would result in greater ranges from start to finish of either drill – however, more data needs to be collected to confirm this correlation.

CONCLUSION

The study is the first to evaluate mobility and efficiency among wheelchair tennis athletes using 3D motion capture modeling. Findings from this approach to sport performance analysis are useful to coaches looking to develop more aggressive court play among wheelchair tennis athletes, as the Pyramid drill resulted in athletes shifting their ball contact speed to earlier in their swing phase. This more aggressive play equates to quicker returns and decreased time given to opponents to adjust to the return. Additionally, athletes with experience in the Pyramid drill showed significantly lower changes (range) in energy expenditure, possibly conserving energy during match-similar drills. 3D motion capture modeling has much to offer to coaches of adaptive athletes, and should be a regular part of any training program. This study developed the first wheelchair athlete 3D model, from which future research and athlete evaluation can emerge.

DISCUSSION

Biomechanical analysis has endless opportunities to deepen our knowledge of sports science and orthopedic and neurological rehabilitation. Three-dimensional motion capture is widely used to assess kinematic and kinetic variables with greater capabilities and precision than manual measurement assessments. Limitations to this approach include its cost and ease of use. However, as technology improves the efficiency of 3D motion capture data collection also improves. On the cusp of this research field, is the need to quantify movements of populations utilizing alternative forms of therapy and recreation. These populations are already at a biomechanical disadvantage, and the need to quantify their movements is more imperative to ensure safety, improve rehabilitation practices, and optimize assessments.

Within Chapter 1 of this dissertation, a 3D human model was created and integrated with force plate and EMG integration to create a robust and portable gait assessment system. Three patients were analyzed individually and kinematic results exhibited similar movement patterns for pelvic tilt, obliquity, rotation, and hip flexion/extension to those found in literature (Baker, 2001b; Lewis et al., n.d.). Each patient expressed improvements in gait symmetry and velocity, and some improvements were statistically relevant. The correlation between hip flexor strength and gait velocity (Hsu et al., 2003; Maria Kim & Eng, 2003; Nadeau et al., 1999) was inferred to be the reason for significant improvements in hip flexion/extension and, consequently, gait velocity. While additional data is needed to draw this conclusion, this study provided quantitative outcomes for inpatient rehabilitation and better determine patient recovery timeline.

The unique nature of stroke rehabilitation is the primary focus on retraining gait patterns and the focus on the patient’s pelvis. In a standard rehabilitation setting, most patients cannot achieve greater kinematic stimulation or muscular activation than they can achieve on their own. As an alternative form of rehabilitation therapy, equine assisted activities and therapies (EAAT), or hippotherapy (HPOT), is sought after to induce gait-similar movements on a rider and stimulate muscular activity that they cannot produce in a standard rehabilitation setting. Chapters 2-5 utilize a wide breadth of data collection methodologies to create an inclusive exploration of the physical and psychosocial effects of horseback riding.

An initial study (Chapter 1) discussed the two-dimensional peak range values and visible representation of horse and rider hip flexion/extensions. Results of equine gait patterns and values were supported by similar findings in literature (Byström, Rhodin, von Peinen, et al., 2009b; Eckardt & Witte, 2017b; Goodworth et al., 2018b; Münz et al., 2013b; Wang et al., 2018b) More, the double pitching pattern of the rider’s pelvis-femur joint, induced by the four-beat gait pattern of the horse’s gait, was also similar to values in literature (Münz et al., 2013b; Wang et al., 2018b). Even more, the study found significant improvements in the administered balance assessment pre and post eight weeks of horseback riding activities. More, in Chapter 2, the same participant population expressed percieved benefits in pyschosocial well-being (including QOL, stress relief, positive emotions, self-efficacy, motivation, and social comparison), and physical well-being (including gait and balance confidence). Other studies have reported older adults experienced positive memories, enjoyment of being outdoors, (K. Lee et al., 2019) and enhanced emotional well-being (Fields et al., 2018). The interest in quality of life, stress relief, positive emotions, self-efficacy, motivation, and social comparison in the older adult population following an EAA intervention.

Further investigation (Chapter 4) addressed the similarities in muscular activation between horseback riding and healthy human gait. Very few significant differences were found, which may have useful implications for supporting EAAT or HPOT as an alternative form of rehabilitation. More, the erector spinae (experienced riders, horseback riding) and left rectus abdominus (novice riders, horseback riding), exhibited significant significantly greater peak contractions. It may be promising to see this significant difference because horseback riding may promote greater trunk muscular activation and, potentially, decrease asymmetrical musculature in riders (M. J. Kim et al., 2018; Y. N. Kim & Lee, 2015; McGibbon et al., 2009). More, this dissertation expanded on a three-dimensional motion capture model to address the kinematic and kinetic relationship between horse and rider (Chapter 5). The three-dimensional rotations of the rider’s pelvis are believed to be directly linked to the rotations of the horse’s pelvis and thorax (Byström, Rhodin, von Peinen, et al., 2009b; Clayton & Hobbs, 2017; Münz et al., 2014a), and this chapter found significant relationships between the magnitudes of rotation (flexion/extension and rotation) of the horse’s lumbosacral joint and the rider’s pelvic responses. The strong positive correlation between the horse’s lumbosacral flexion/extension and the rider’s hip joint flexion/extension was evident. However, due to the horse's thoracic rotations during gait, the greater rotation of the horse's pelvis may elicit a greater stabilization response by the rider's lower limbs and decrease the flexion/extension of the rider's hip joints. A larger data set would provide greater insight to this theory and the effect of the lumbosacral joint rotations on rider's hip joint moments. These relationships followed similar trends and have the potential to indicate muscular activation based on forces acting on the rider's hip joint, and could ultimately have implications in choosing horse types for use in EAAT or HPOT.

Because of the double pitching pattern in pelvis-femur flexion/extension and the magnitudes of ranges of motion and joint moments, horseback riding (EAAT, HPOT) could have significant implications for patient rehabilitation. When comparing pelvis patterns (Figure 42), it is obvious that the human receives similar two-dimensional stimulation while horseback riding as they create while walking. However, while horseback riding, the rider experiences two pitching cycles for a single stride of the horse, in contrast to the single pitching cycle during human gait. This stimulation while horseback can be continuous without putting the strain and weight on the participant's lower limbs like human ambulation. In further support, range of motion values found in pelvic tilt and hip flexion/extension during inpatient stroke recovery (P3) and while horseback riding was relatively similar, depending on the horse (Table 6 and Table 14). These results support the belief that EAAT and HPOT could provide equivalent levels of stimulation for lower limb and trunk rehabilitation for participants will limited mobility.

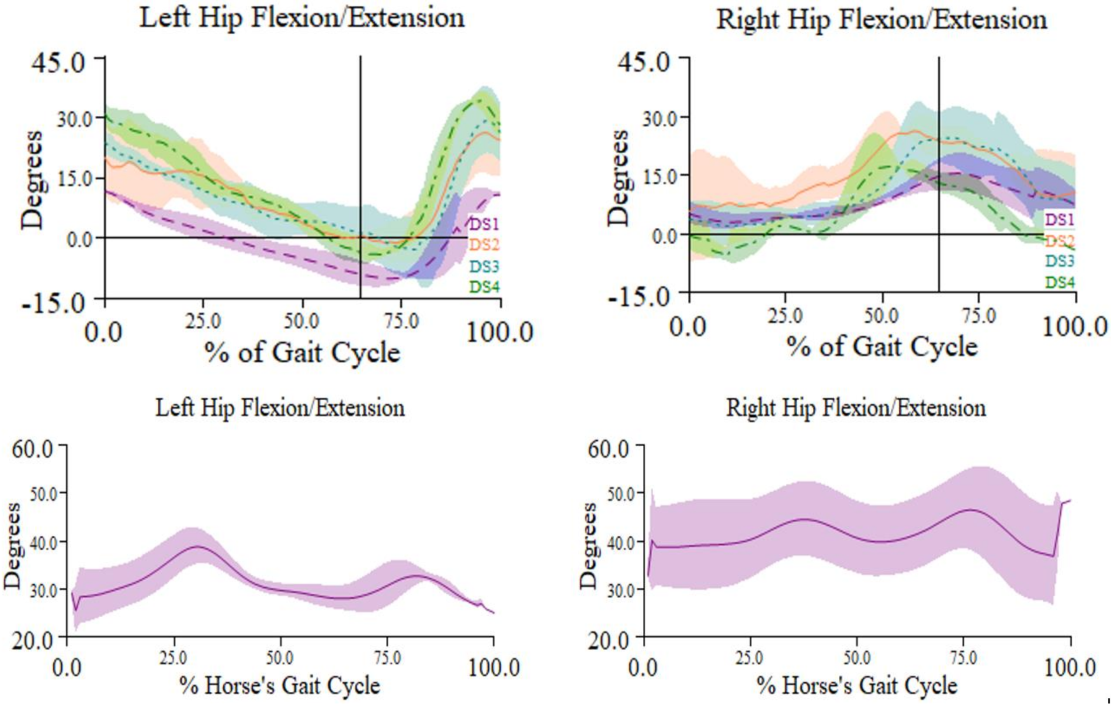

Figure 42. Comparison of human gait (top) and horseback riding (bottom) hip flexion/extension

Table 6. Quantitative values for range of motion in the desired kinematic assessments. Statistically significant differences were present between DS1 and DS2 (*) and between right and left measurements (**).

Data Set	Pelvic Tilt	Pelvic Obliquity	Pelvic Rotation	Left Hip Flexion/Extension	Right Hip Flexion/Extension
DS1	2.46±0.82	5.17±1.97	6.85±0.82	23.72±2.27*,**	12.67±2.56*,**
DS4	3.94±4.70	3.77±1.01	5.20±0.66	37.44±4.38*,**	21.85±2.93*,**

Table 14. Kinematic and kinetic variable results following data processing. Values are total ranges of motion expressed by rider and horse joints during 0-100% of the horse's gait cycle.

	α	γ_L	γ_R	$M_{X,L}$	$M_{Y,L}$	$M_{Z,L}$	$M_{X,R}$	$M_{Y,R}$	$M_{Z,R}$	σ	β	L
	Degrees	Degrees	Degrees	N*m/kg	N*m/kg	N*m/kg	N*m/kg	N*m/kg	N*m/kg	Degrees	Degrees	m
Zues	2.186	20.084	22.505	0.223	0.661	0.025	0.284	0.430	0.064	9.200	4.462	0.927
Mallorie	8.411	20.061	14.613	0.169	0.186	0.037	0.000	0.000	0.000	7.738	10.899	0.914
Sierra	3.681	13.776	15.689	0.173	0.156	0.019	0.242	0.285	0.027	6.509	7.140	0.883
Junior	4.412	9.621	11.030	0.140	0.129	0.015	0.200	0.189	0.034	5.784	9.633	0.876
Brody	10.605	17.777	18.075	0.187	0.140	0.027	0.475	0.868	0.170	0.000	0.000	0.864
Ranger	6.028	19.599	15.900	0.147	0.089	0.039	0.283	0.230	0.025	8.073	6.284	0.800

Even more, once disabled populations are able to return to their activities of daily living, they still face a biomechanical disadvantage in recreation to maintain their health and exercise. Within Chapter 6, a novel 3-dimensional wheelchair athlete model was created for kinematic and kinetic analysis This model was specifically made to assess wheelchair tennis athletes, however, it can be adapted to other wheelchair sports. This model successfully measure relevant shoulder joint kinematics, racquet velocity (Landlinger et al., 2010) and energy expenditure. Athletes with no prior experience in the Pyramid drill exerted more energy in the new drill than the traditional Figure 8 drill. However, athletes with familiarity in the Pyramid drill were more energy efficient and this could represent greater energy conservation during turns. Because this drill simulates match-like situations, it would be beneficial for athletes to find the most energy efficient approach to the drill, however, additional data is needed to draw a conclusion.

CONCLUSION

This dissertation addressed the usage and benefits of three-dimensional motion capture modelling and its ability to increase quantitative knowledge of rehabilitation and recreation for disadvantaged population groups. Specifically, this dissertation focused on 3D models and movement profiles for human gait analysis with emphasis on post-stroke patients, with direct model translation to analyze equivalent measurements while horseback riding in use of the alternative form of rehabilitation, equine assisted activities and therapies or hippotherapy. More, the same technology and data collection approach was used to assist in the 3D model development and movement science assessment of wheelchair tennis athletes. The model development and kinematic and kinetic analysis methodologies were similar and translated through chapters of this dissertation. The quantitative results, coupled with the qualitative assessment in Chapter 3, provide a robust assessment of the effects of 3D movement analysis on rehabilitation and adaptive activities.

FUTURE WORK

As previously stated, larger data sets would draw more conclusive findings to these assessments. Three additional patients will be recruited for the study in Chapter 1 and strengthen the results and potential findings. More, additional data on the horse and rider relationship will continue and potentially have greater implications on the EAAT and HPOT communities. The biomechanical assessment of wheelchair tennis athletes was only the cusp on research within this field, and future work is being pursued with this group. Future work for all assessments will incorporate deeper understanding of patient's, rider's, and athlete's muscular activity to deepen the quantitative modelling and interpretation of the biomechanical assessments for these population groups.

APPENDICES

Appendix A

Individual ranges of motion in right and left hip flexion/extension.

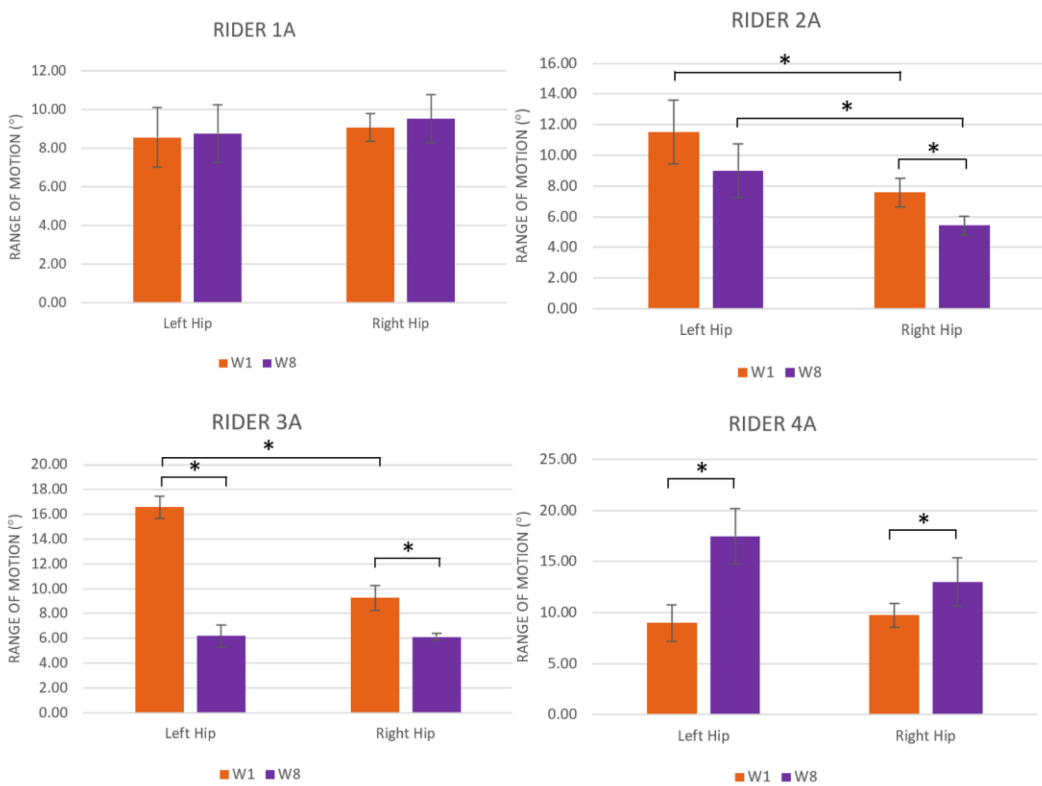

Figure 43. Hip flexion/extension range of motion values for horseback riders on Horse A. Statistically significant differences in values are represented by *.

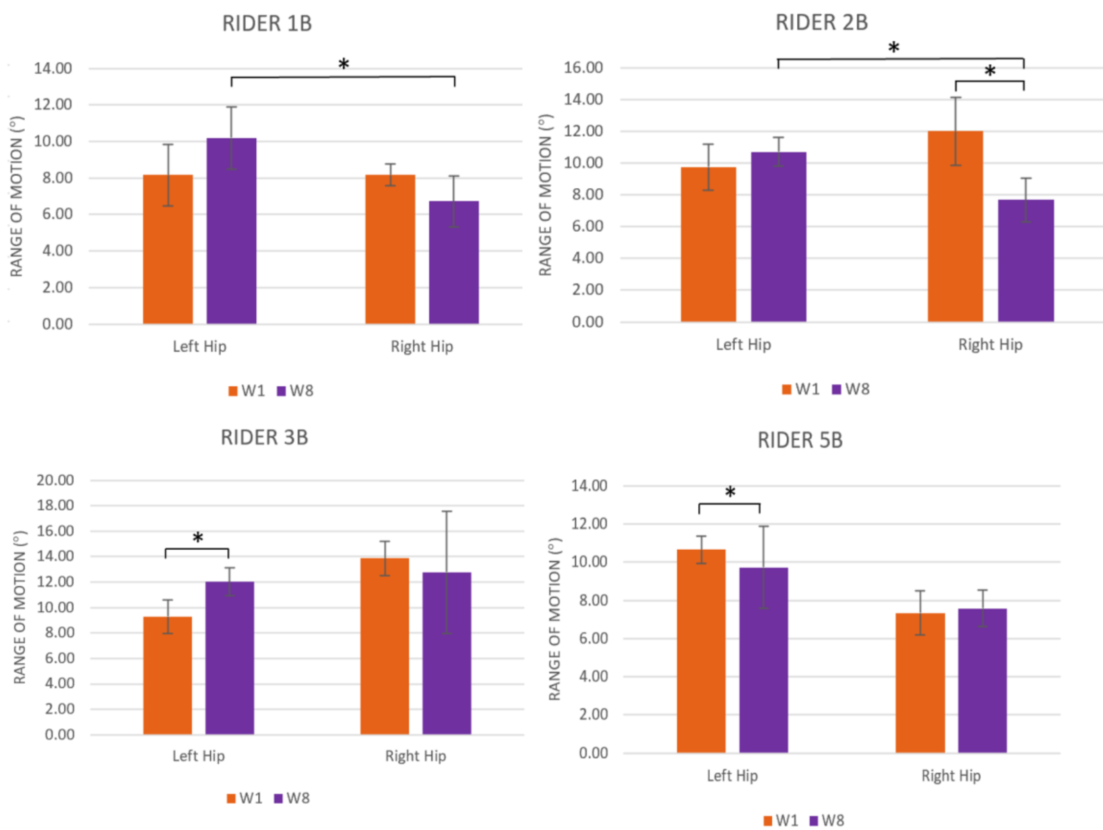

Figure 44. Hip flexion/extension range of motion values for horseback riders on Horse B. Statistically significant differences in values are represented by *.

Appendix B

Individual kinematic waveform reports for wheelchair tennis athletes

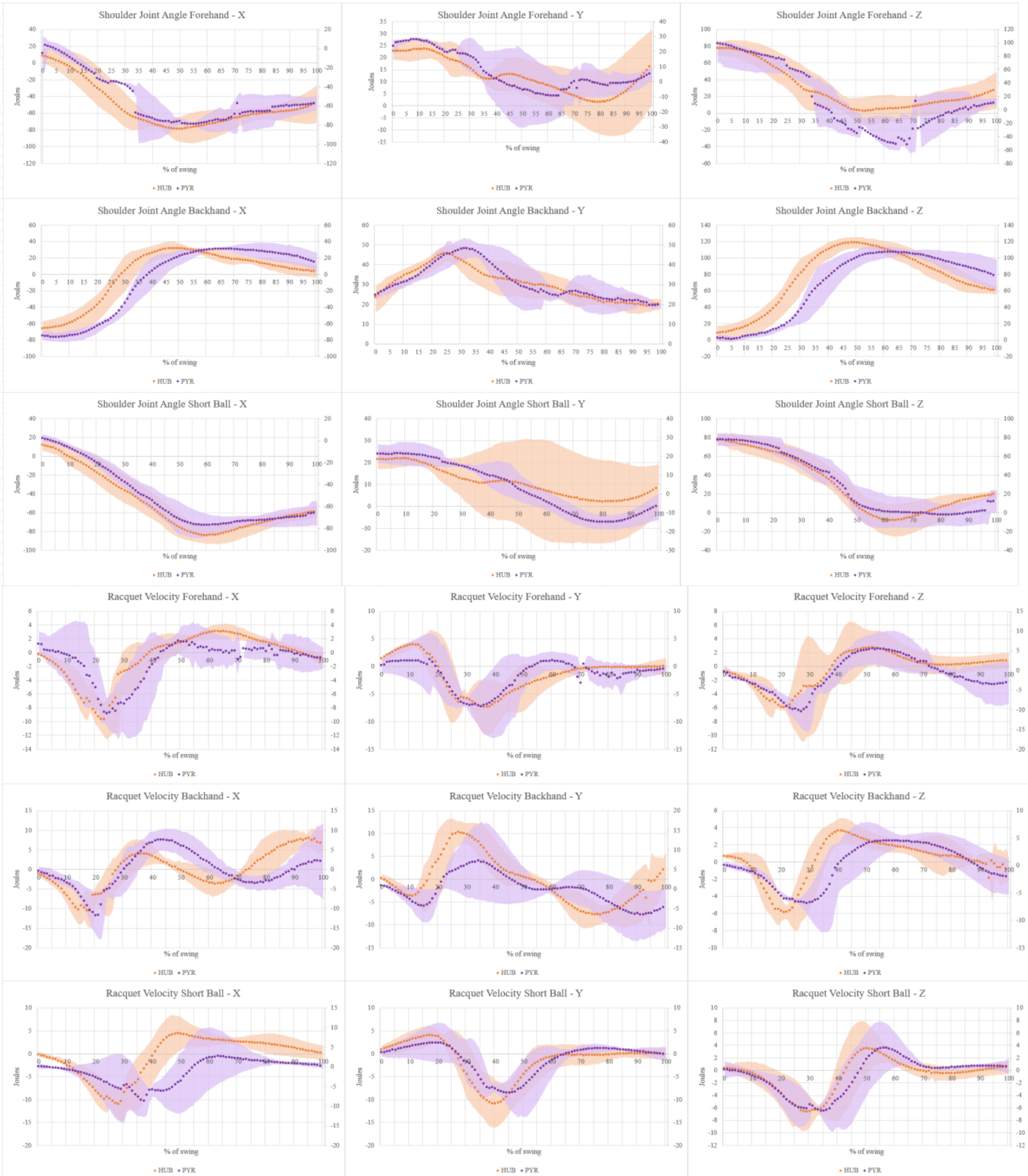

Figure 45. Athlete 1 waveforms for shoulder joint angles and racquet velocity for the Pyramid (purple) and Figure 8 (orange) drill.

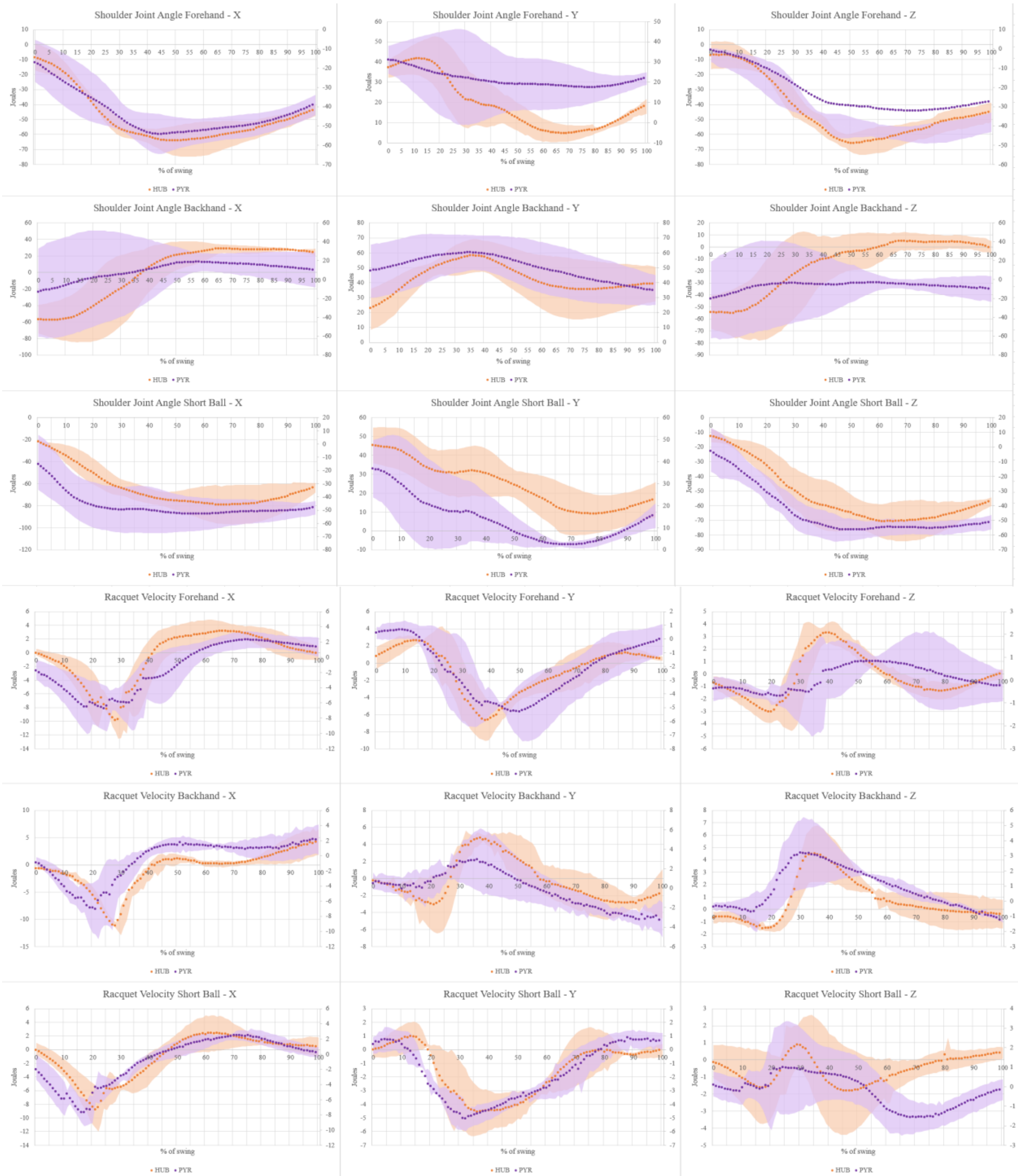

Figure 46. Athlete 3 waveforms for shoulder joint angles and racquet velocity for the Pyramid (purple) and Figure 8 (orange) drill.

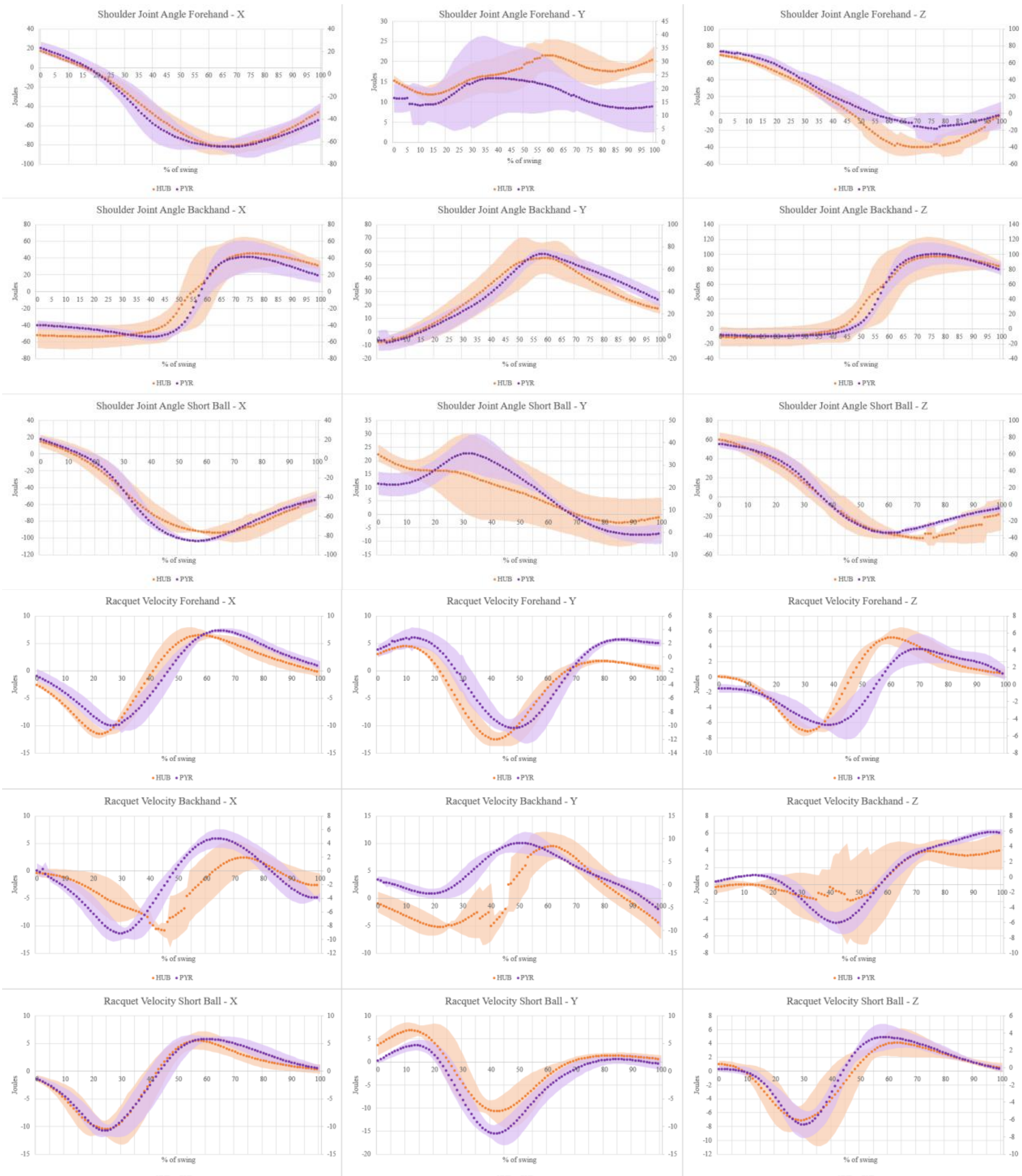

Figure 47. Athlete 4 waveforms for shoulder joint angles and racquet velocity for the Pyramid (purple) and Figure 8 (orange) drill.

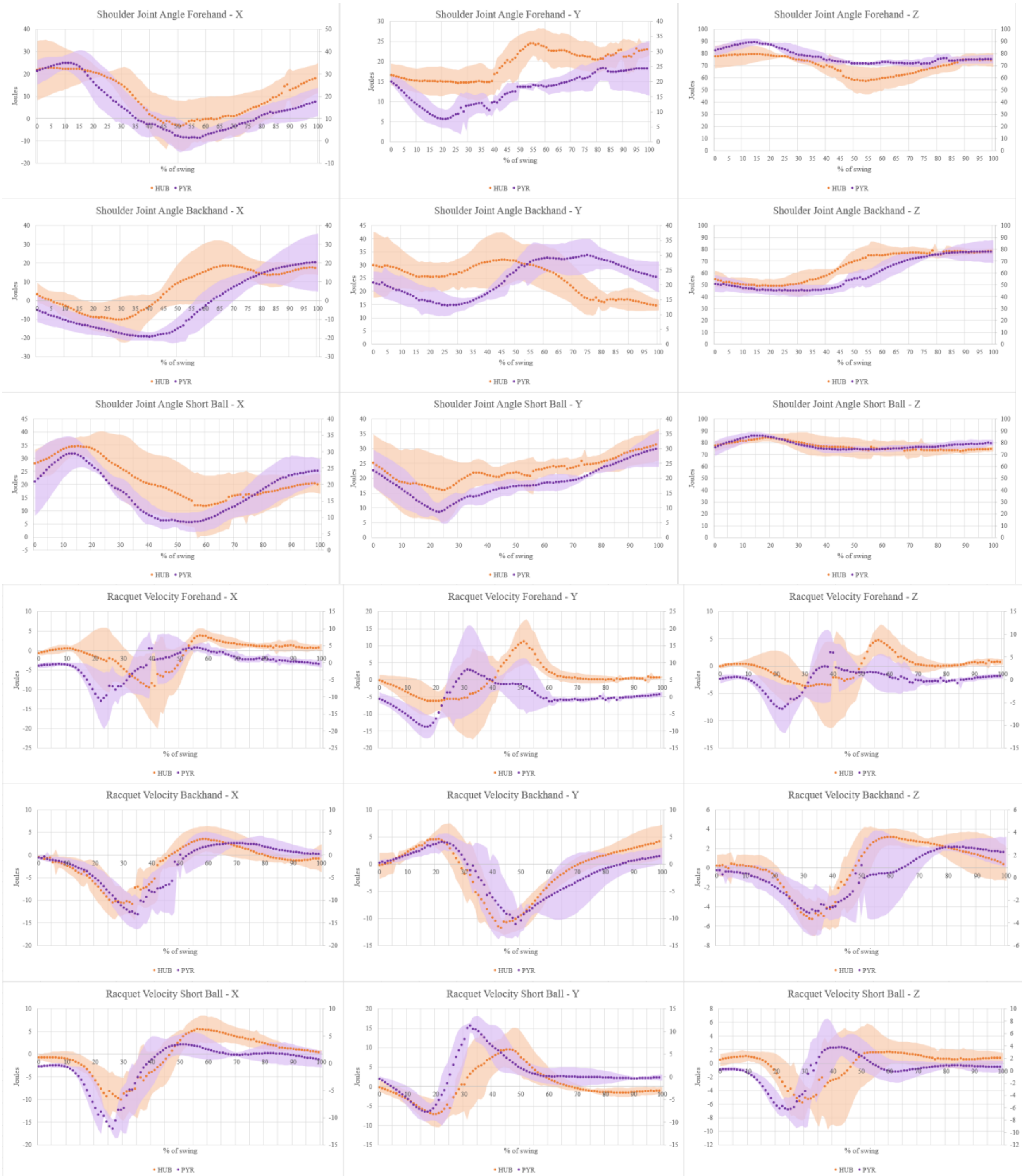

Figure 48. Athlete 5 waveforms for shoulder joint angles and racquet velocity for the Pyramid (purple) and Figure 8 (orange) drill.

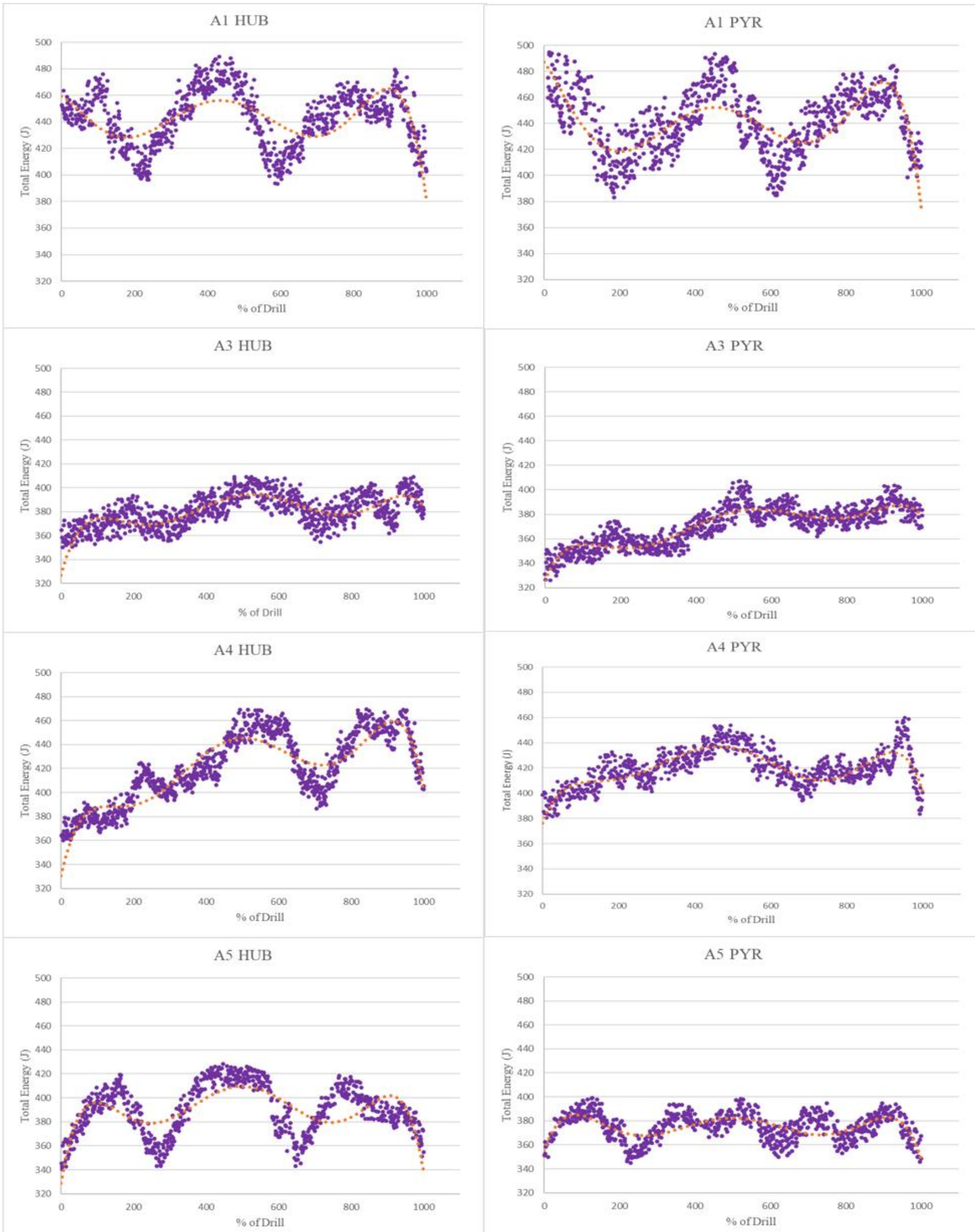

Figure 49. Total energy values from 0-100% of the Figure 8 (left) and Pyramid (right) drills.

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