Therapeutic Intervention or Rider Influence on the Electromyographic Activity And Kinematics of Horses

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DEDICATION

This research is dedicated to the horses in my life who have given so much to me, and who have inspired me as a veterinarian to make the lives of other horses better.

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I want to thank all of the individuals who supported me throughout the process of completing this thesis. Particularly my mentors, collaborators, and co-researchers; Dr. Tena Ursini, Dr. Steve Adair, Dr. David Levine, Dr. Jim Richards, and Dr. Songing Zhang—each of whose insight, training, and patience was appreciated and invaluable for the completion of this research. I also want to thank all of the individuals who helped with the data collection and horse handling (with lots of running); Emma Horn, Dawn Phillips, Josh Bowlin, Tyler Moore, Lauri Terrones, Lindsey Pokrzywinski, Kylie Durham, and Cailey Nichols. I am also grateful for our wonderful VREC teaching herd of horses, who participated in this study.

PREFACE

Equine sports medicine and rehabilitation is an evolving field that has seen much growth in the last decade. Similar to trends observed in human sports medicine, there has been a shift in paradigms with a greater emphasis placed on injury prevention and maximizing performance through a deeper understanding of biomechanics and the kinetic chain. Veterinarians commonly intervene and initiate treatment on horses following the development of an injury. However, if veterinary and equine medical professionals could begin to play a bigger role in recognizing a horse's biomechanical faults, functional restrictions, or predisposition to injury, then perhaps we could have a more pro-active and preventative role towards injury in our equine athletes. Or additionally, following an injury, we can use that same biomechanical knowledge to maximize healing of an injury and prevent secondary compensatory problems related to the primary injury. To maximize care of our equine athletes, we need to continue researching and pushing the boundary of knowledge forward, making advancements in this area through an integrated biomechanical approach.

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INTRODUCTION

Postural Control And Equine Biomechanics

Human physical therapy has grown since the early 20th century and has become an integral part in the healthcare team with an emphasis on the maintenance, restoration, or enhancement of physical and functional abilities of the individual [1,2]. The principles and techniques used in human physical therapy have a strong evidence base, whereas in the equine field the scientific evidence supporting therapeutic exercise strategies is lacking. For this reason, there has been a cross-species application of many physical therapy principles. Most therapeutic exercise strategies used in horses involve the application of the principles, knowledge, and experience in human physical therapy, combined with the understanding of equine biomechanics and movement dysfunction associated with injury [1,3-10]. However, to better understand the physiological effects of these therapeutic interventions, further research is needed to support their use.

Maximizing performance and the biomechanical efficiency of movement in equine athletes often involves posture control and core muscle engagement [1,9-10]. A state of good posture has been previously defined as “a state of musculoskeletal balance that protects the supporting structures of the body against injury or progressive deformity” [2]. Posture control and core muscle strengthening is encouraged to avoid limb overload in horses recovering from injuries to their extremities, as reduced core strength and poor posture can lead to increased loading of an injured limb [1,9-10]. Posture is influenced by many factors including conformation, body condition, muscle development, muscle hypertonicity, and muscle atrophy [1]. Spinal positioning and posture control are influenced by several muscle groups including the; abdominal core muscles, hindquarter muscles, epaxial, and hypaxial muscles, which have been shown to play a significant role in movement and stabilization of the spine throughout the gait cycle [9-18]. The muscles investigated in this study were the longissimus dorsi (LD) and rectus abdominus (RA) given their contributions to core and postural stability.

Therapeutic techniques that have been purported to improve postural control in horses include assistive or adaptive equipment that provide tactile and proprioceptive stimuli, whose sensory input is integrated into functional activity. One such technique is the application of elastic resistance bands in a figure of eight conformation around the horse’s trunk to provide a rhythmic stimulus to induce hindquarter protraction through suspected simulation of the iliopsoas and spinal flexor muscles [1]. Alternatively, elastic resistance bands have been applied from the girth of a saddle or surcingle, extending around the caudal aspect of the hindlimbs to facilitate hindlimb protraction and engagement of the core musculature [1]. A 4-week training regimen using elastic resistance bands around the abdomen and hindquarters resulted in alterations of back kinematics as measured by inertial measurement unit (IMU) sensors [3]. Changes seen

included a reduction in mediolateral and rotational movement of the thoracolumbar region at a trot, consistent with increased dynamic stability of the spine [3]. While alterations in back kinematics have been identified with the use of an elastic resistance band training system, changes in activation of epaxial muscles of the spine, such as the LD muscles, have not yet been investigated. Additionally, while an increase in activation of abdominal musculature with the use of the abdominal elastic resistance band has been suggested, changes in core abdominal muscle recruitment with electromyography have not yet been investigated to the authors' knowledge.

Negotiating over ground poles or raised ground poles (cavalettis) is another therapeutic technique used in rehabilitation programs for training proprioceptive awareness, core muscle activation, posture development, and neuromotor control [1,5,19-22]. In addition to an altered neuromuscular response, maneuvering over obstacles can also result in a learned response, which can influence foot clearance, joint trajectories, and electromyographic activity [19,23]. The use of raised poles incorporated into a gymnastic training program has been shown to increase stride length, tracking distance, and improve multifidi muscle cross sectional area when combined with dynamic mobilization exercises [4]. While the basic locomotor rhythm in horses is mediated at the spinal level by the central pattern generators, maneuvering over obstacles requires visual input and processing in the motor cortex to produce an altered locomotor response [24]. The locomotor response initiated with ground poles includes changes in limb elevation and flight arc, increased swing phase joint flexions of the forelimbs and hindlimbs, and increased lateral stability of the stance limb likely from recruitment of the forelimb adductor and extensor musculature [5, 19-20]. Additionally, as a horse maneuvers over an obstacle such as ground poles, their balance is perturbed as the swing limbs are raised over the obstacle, which requires the body and stance limbs to maintain balance and stability [10]. Identifying therapeutic strategies to increase muscle activation of the trunk is important for increasing spinal stability by limiting the degree of spinal rotation [25].

The LD muscle was investigated in this study, as the thoracic segment is the largest of the epaxial muscles, which has a significant role in trunk stability [10,26]. The RA muscle works in accordance with the LD to generate tension and dorsal flexion in the thoracolumbar spine, helping to create a stable and rigid platform to facilitate locomotion [10,18,26]. Given the increase in forelimb and hindlimb flight arc and joint angles with ground poles and the resulting increase in spinal rotation, our hypothesis was that there would be increased LD and RA muscle activation when walking and trotting over ground poles given its role in stabilizing the axial skeleton. A significant increase in mean and maximal surface electromyographic activity of the RA muscle has been reported in 3 out of 6 horses when ridden over ground poles and raised poles [11]. Further research is needed to investigate and better characterize changes induced in muscle

activation of core, epaxial, hypaxial, forelimb and hindlimb muscles with the use of ground poles at a walk and trot.

Electromyography

Electromyography (EMG) is a technique used to evaluate muscle activity during movement, through the measurement of electrical activity at the level of the muscle fibers [27-32]. Skeletal muscle, which is involved with voluntary movement, is innervated by a type of motor nerve called an alpha motor neuron. An alpha motor neuron has its cell body located in the ventral horn of the spinal cord, with terminal branches innervating multiple skeletal muscle fibers [27,30]. The functional unit consisting of one alpha motor neuron, the neuromuscular junction, and the muscle fibers it innervates is considered a motor unit. A motor unit is the smallest controllable muscular unit which produces a depolarization wave called a motor action potential (MAP), which propagates in both directions along each muscle fiber from the motor end plate [30]. Muscle fibers have a specialized type of excitable semi-permeable membrane called the sarcolemma, which upon stimulation by the alpha motor neuron results in membrane depolarization immediately followed by a repolarization phase. This depolarization-repolarization cycle forms a depolarization wave or electrical dipole which travels along the surface of the muscle fiber and can be detected from the surface of the skin [27,30]. Surface electrodes placed over superficial muscles of interest are able to detect the summation of action potentials across several muscle fibers within the vicinity or “pick-up” zone of the electrode pair. This summation of individual muscle fiber action potentials detected by the electrode pair is termed the motor unit action potential (MUAP), which is amplified to produce the raw EMG waveform [30]. The raw EMG signal can be processed to remove artifact and to enable comparisons of muscle activity within and across individuals [25,27-30].

Numerous EMG studies have been conducted in the horse in both clinical and nonclinical environments. Key areas that have been researched include a muscles role in pathologic conditions, and the activity of muscles involved in locomotion to determine their contribution to performance [11,14-15,18,27]. EMG has been utilized to quantify muscle activity within defined events to support riding or training practices in horses such as the use of gymnastic gridwork, interval training in racehorses, riding with hyperflexion, or the use of specific training aids [27].

Surface Electromyography

Surface electromyography (sEMG) involves the placement of a non-invasive surface electrode over the superficial muscle of interest [27,30,32]. The myoelectrical activity of certain muscles can be measured to determine changes in muscle activation during defined events, during the use of therapeutic devices, or as a result of specific training or rehabilitation strategies [27]. Other applications for sEMG in horses include the evaluation of baseline muscle activity

in static situations, the investigation of neuromuscular dysfunction, and the study of dynamic muscle recruitment patterns, adaptation to training, or the result of specific interventions on muscle activation [27].

Data Processing

Processing of sEMG data is performed to remove noise artifact from the raw sEMG signal and to acquire useable data for further analysis [27-32]. While no set methods have been accepted as gold standard for equine sEMG analysis, typical approaches to processing sEMG data include applying filters to ensure only relevant frequencies that contribute to the event are being analyzed [27]. Filters involve a cutoff frequency to remove all frequencies below or above a certain cutoff point [25,27-28]. A low-pass filter can attenuate higher frequencies to minimize "noise" components from the signal. A high-pass filter can attenuate lower frequencies to minimize baseline "noise" and movement artefact. Applying a band pass filter would be equivalent to applying both a low-pass and high-pass filter, to remove low and high frequency artifact from the signal.

Other types of signal processing involve full-wave rectification which converts negative amplitudes to positive amplitudes, by reflecting all negative amplitudes about the baseline [30]. This enables standard amplitude parameters like mean, maximum, or area under the curve to be evaluated. Another important sEMG signal processing step is amplitude normalization, which is a means of calibrating the signal microvolts value to a designated calibration unit so that the data can be scaled relative to the selected reference value [25,27-28,30]. Given that the signal values are largely influenced by the detection condition, normalization methods can eliminate the influence of this variable and allow comparison of data which is scaled to the reference value. This reference value can be a maximum voluntary contraction (MVC) or a reference voluntary contraction (RVC). MVCs are obtained when individuals perform specific maneuvers to maximally contract a particular muscle group, which is commonly performed in humans. RVCs often involve the maximum sEMG outcome detected across all conditions for a particular individual at a particular gait, which are utilized for normalization more frequently in animals, pediatrics, or with human participants experiencing pain or neurological disorders [27-28].

When examining differences in muscle function during gait, two commonly reported amplitude-based outcome measures in equine sEMG literature include integrated EMG (iEMG) and average rectified value (ARV) [28]. The iEMG value involves integration of the sEMG signal over a specified time interval and represents the area under the voltage curve. The ARV represents the mean value of the full-wave rectified sEMG signal which is measured over a specified time interval [28]

Kinematics

Locomotion consists of a complex coordination of movements of the body and limbs consisting of both translational and rotational motion of various body segments [33]. Kinematics is the area of study that deals with describing the motion of bodies without considering the forces that cause motion to occur [34]. Kinematics is a fundamental component of gait analysis as it involves obtaining quantitative objective information about the motion of the individual. In horses, gait analysis is important for understanding the normal motion of healthy individuals, as well as for characterizing changes in gait pattern induced by injury or pathology. Additionally, changes in gait pattern are often influenced with therapeutic or training techniques, which can be better understood with kinematic gait analysis systems.

To quantify movement of the horse, certain noninvasive techniques have been developed utilizing markers fixed to the skin over predefined anatomic landmarks [35]. The type of information which can be obtained with a kinematic gait analysis system includes joint range of motion, joint or segment velocity or acceleration, as well as angular motion. Kinematic analysis relies on two basic assumptions; that all motion segments will behave like rigid bodies, and that the skin markers reflect movement of the underlying skeleton [33-35]. Through kinematics, the movement of individual body segments can be described as they move and rotate relative to each other around axes located at the joints [33]. Based on the positioning and number of skin-fixated markers used, it is possible to perform analysis in a 2-dimensional (2-D) or 3-dimensional (3-D) approach. With 2-D analysis, data is analyzed on one plane, such as the sagittal, horizontal, or frontal plane. With 3-D analysis, out-of-plane or rotational motions can be investigated [35]. The angular motions of joints are described in three directions, which include; flexion and extension, abduction and adduction, and internal and external rotation.

The kinematic analysis of horses presents with certain challenges, such as the presence of large superficial muscle groups overlying certain parts of the skeleton. Additionally, increased skin elasticity in horses may contribute to increased motion artifact. When evaluating the horse and rider interaction, certain limitations exist in investigating spine kinematics, given the position of the rider over the thoracic spine [35]. However, despite these limitations, kinematics remains a useful area of study for the investigation of equine stride, gait, body segment, and joint characteristics.

CHAPTER I
EQUINE ELECTROMYOGRAPHY WITH THERAPEUTIC EXERCISE

A version of this chapter was submitted to a peer-reviewed journal but has not yet been accepted for publication.

This article was revised by mentor and collaborator Dr. Steve Adair. Dr. Tena Ursini was a co-researcher in the data collection, processing and interpretation of the data presented in this article. Jim Richards was a collaborator who assisted with the data processing, statistical analysis, and interpretation of the findings presented in this article.

Abstract

Core strengthening and postural stability are desired outcomes of certain therapeutic exercises performed in horses. This study aimed to quantify changes in muscle activation at a walk and trot in horses traveling over eight consecutive ground poles evenly spaced (at 30 inches for walk and 48 inches for trot) in parallel fashion in a straight line, and with hindquarter and abdominal elastic resistance bands applied at 25% stretch. Surface electromyography (sEMG) data were collected for the longissimus dorsi and rectus abdominus muscles in six horses. A 2x2 repeated measures ANOVA was performed for each muscle to test for significant differences in differences in normalized average rectified values and maximum low pass signals. Within subject effects were reported, followed by post-hoc pairwise comparisons to evaluate differences between the conditions of with or without ground poles or elastic resistance bands. The use of ground poles at a walk resulted in a significant ($p < 0.05$) increase in the maximum low pass value bilaterally in the longissimus dorsi and rectus abdominus muscles, with an increase in the average rectified value bilaterally in the rectus abdominus muscles and right longissimus dorsi muscle. The use of ground poles at a trot resulted in a significant increase in the maximum low pass value bilaterally in the rectus abdominus muscles. The hindquarter and abdominal elastic resistance bands resulted in a respective 27% and 27.2% increase in the mean average rectified value of the left and right RA muscles, however this only reached statistical significance in the left RA ($p < 0.05$). These findings provide support regarding changes in muscle activation when using ground poles to increase core and epaxial muscle engagement. While a significant effect on core muscle activation was identified with the elastic resistance bands at a trot, further research is needed in this area to further characterize their effects on muscle activation.

Aim

The main purpose of this study was to evaluate the mean average rectified value (ARV) and maximum low pass value of the LD and RA muscles at a walk and trot with two therapeutic interventions: with the use of ground poles, and with the use of elastic resistance bands. For the purposes of this study, eight

consecutive ground poles were arranged in parallel fashion in a straight line at specific distances to accommodate stride length differences for a walk and trot (0.76m for walk, 1.22m for trot). Small adjustments in pole spacing (± 0.07m) were additionally made to accommodate for individual horse variation in stride lengths at either gait [5,19]. Elastic resistance bands were applied behind the hindquarter musculature and ventral abdomen, and were applied at 25% stretch. Horses additionally were taken through the ground poles with and without the presence of the resistance bands, with data investigated for an interaction between the poles and resistance bands. The specific muscles selected were based on their functions related to posture control [12-13,25-26,36]. Earlier research has demonstrated the activity of these muscles during locomotion at a walk and trot, without the presence of therapeutic intervention [14-18].

Our aim is to assess whether the use of ground poles or a proprioceptive aid provided by hindquarter and abdominal elastic resistance bands would result in differences in surface electromyographic activity based on normalized ARV and maximum low pass values of selected muscles at a walk and trot, and to determine if an interaction between the conditions existed. We hypothesized that the LD and RA muscles would have increased sEMG activity, characterized by increased normalized ARV and increased maximum low pass values, when ground poles and elastic resistance bands were used.

Materials And Methods

Horses

This study was approved by the University of Tennessee Institutional Animal Care And Use Committee (IACUC). Six horses were included in this study. There were 5 mares and 1 gelding, aged between 3 and 20 years old, and with 4 different breeds or crosses represented (3 Thoroughbreds, 1 Quarter Horse, 1 Tennessee Walking Horse cross, 1 Saddlebred cross). All horses were trotted in hand and evaluated by two veterinarians experienced in lameness. Any horse with lameness on hard or soft ground greater than grade 1 on the AAEP scale was excluded. Horses that were gaited without diagonal two-beat trot or with apparent neurological deficits were also excluded. All selected horses were exposed to the experimental set-up (arena, ground poles), trained to lunge with a surcingle, and were instrumented with elastic resistance bands prior to data collection. The acclimation period to the elastic resistance bands ranged from 1 to 4 days, depending on each horse's level of earlier training or exposure. Throughout the acclimation and data collection period, subjects were administered a low-dose non-steroidal anti-inflammatory drug (phenylbutazone, 2.2mg/kg, once daily) to reduce the risk of exercise-induced soreness in the current sample population of horses that are not in a routine exercise program. Since subjects were screened and excluded for consistent lameness or asymmetry, low-dose non-steroidal anti-inflammatory drug is expected to result in minimal alteration of the gait pattern of the included horses.

Equipment And Instrumentation

Each horse was tacked with a modified saddle pad with the caudal thoracic region removed to prevent interference with the LD sensors. The saddle pad had buckles for attachment of the abdominal and hindquarter elastic resistance bands. A breast collar and surcingle were used to hold the saddle pad and resistance bands in place (Figure 1). Elastic hindquarter and abdominal bands (Equicore Concepts LLC, East Lansing, Michigan, USA) were fitted with a 25% stretch, which was calculated from 75% of the distance between attachment points of the elastic bands. There was not obvious deformation or indenting of the hamstring or abdominal musculature after the bands were applied.

For sEMG electrode placement, the skin over the muscles was shaved, then cleaned with chlorhexidine scrub and isopropyl alcohol. Pre-gelled bipolar self-adhesive Ag/AgCL electrodes (HEX Dual Electrodes, Noraxon, Scottsdale, Arizona, USA) with an inter-electrode distance of 2cm, were placed parallel to the direction of the muscle fibers in defined locations and secured with cyanoacrylate adhesive. Electrodes were manufactured with a fixed 2cm inter-electrode distance and unable to migrate (Figures 2 & 3). This inter-electrode distance falls within the range of what is utilized in other equine sEMG studies (ranging from 1 to 4cm inter-electrode distance) and is based on the European SENIAM guidelines [14,16,18-19,24-25,32]. Wireless direct transmission system (DTS) sensors (DTS Research EMG Probes, Noraxon, Scottsdale, Arizona, USA) were attached to the electrodes with 3-inch pinch end leads and secured to the skin with cyanoacrylate adhesive.

The RA muscles were instrumented with surface electrodes placed mid-length along the muscle bellies bilaterally, 2cm lateral to the midline (Figure 2) [18]. The RA electrode and sensor were located cranial to the abdominal band, to prevent interference of the EMG signal. For the LD muscles, surface electrodes were positioned approximately 2cm lateral to the midline bilaterally over the bulk of the LD muscles at the level of the dorsal spinous process T16 (Figure 3), which was identified with linear ultrasound transducer and marked with liquid correction fluid [14,16,28,39].

Data Collection

All data were collected using the Noraxon MR3.10 software. The sEMG signal was transmitted to the computer by a telemetric system (TELEmyo Direct Transmission System (DTS) Belt Receiver, Noraxon, Scottsdale, Arizona, USA). High speed camera video footage was collected for all trials at 125 frames per second (NiNOX Video Capture, Noraxon, Scottsdale, Arizona, USA), which was synchronized with sEMG, and with digital output (myoSYNC Master Sync, Noraxon, Scottsdale, Arizona, USA). The camera was set up at one end of the arena facing the direct line of movement so that horses traveling away and towards the camera would be included in the field of view.



Figure 1. Placement of the surface electrodes over the LD (arrow) muscles bilaterally, with application of hindquarter and abdominal elastic resistance bands



Figure 2. Placement of surface electrodes over the RA muscles bilaterally, with a fixed 2cm inter-electrode distance and adjacent wireless direct transmission system (DTS) sensors attached to the electrodes with 3-inch pinch end leads.



Figure 3. Placement of surface electrodes over the LD muscles bilaterally at T16. (Needle electrodes were placed simultaneously in the multifidus muscle at the levels of thoracic and lumbar spinous processes T12, T18, and L5. The results of the needle EMG data were not investigated in the current study.)

Following instrumentation, sEMG data were gathered at sampling rate of 1500 Hz with synced video footage. If the movement condition was not met, such as a subjective observation of change in gait, stumbling, bucking, or when sEMG sensors were dislodged, data collection was extended or repeated until a minimum of five consistent three-stride segments were obtained. Typically, the handler would travel the arena length or over the sequence of poles 3 times down and back for the desired strides to be obtained. Data were obtained in-hand at walk and trot, on a straight line at each horse's preferred speed within the gait, following a previously reported method [3,5]. The walking and trot conditions were performed in an indoor arena with soft synthetic surface in the following non-randomized order;

1. Without bands, straight line
2. Without bands, straight line over ground poles
3. With bands, straight line
4. With bands, straight line over ground poles

Eight consecutive ground poles were set up in a parallel fashion in a straight line, spaced to 0.76 $\pm$ 0.07 meters for walking (30 $\pm$ 3 inches) and spaced to 1.22 $\pm$ 0.07 meters for trotting (48 $\pm$ 3 inches). Pole spacing was adjusted by up to 0.07 meters (3 inches) to accommodate the stride lengths of different horses [5,19]. Poles were arranged within an indoor arena on a soft synthetic surface. Ground pole spacing was checked and adjusted every time it was knocked or moved by a horse throughout the data collection.

Validation of Muscle Activation Patterns Across the Gait Cycle

A subsample of the muscle activation data were collected with synchronized kinematic data in order to evaluate the LD and RA muscle activity relative to objectively defined gait events. Four spherical markers (14mm diameter pearl markers, B&L Engineering) were placed on the lateral aspect of the coronary band on all four feet, secured with double-sided adhesive tape and cyanoacrylate adhesive. The kinematic data were collected using a 12-camera 3D motion capture system, with a sampling frequency of 100 frames/second (VICON Motion Systems Ltd, Oxford, UK). Heel strike and toe off events were determined using vertical displacement of the lateral foot markers. These data were used to check the firing sequences with known gait events during walks and trots on hard ground in one of the sample subjects. For one complete gait cycle, two muscle activations were seen in the LD and RA muscles at both the walk and trot, which was consistent with previous reports [12,14,16,25-26]. These trends in muscle activation for each gait cycle were compared and found to be consistent with the strides marked using sEMG synced video footage on soft ground. Since the left hind toe-off gait events for identifying individual strides were marked subjectively based on video footage, this use of motion capture using objectively defined gait events provided a confirmation of the phasic activation patterns for each muscle of interest across the gait cycle, with two complete LD and RA muscle activations per stride. Since the 3D motion capture system was not

available for use during the time of data collection, it was used retrospectively for a validation trial to confirm muscle activation patterns relative to objectively acquired gait events given that the same muscle activation patterns were seen in each muscle irrespective of the surface type. The left hind toe-off event corresponded with a LD or RA muscle activation peak during both walk and trot gaits, whereas the heel strike event proceeded the LD and RA muscle activations.

Data Processing And Analysis

The high-speed camera video footage was reviewed and used to mark a total of fifteen strides (five consistent three-stride segments) for each horse under each trial condition. Three-stride segments were isolated using “tags” at left hind toe-off events (same viewer, KS). Toe-off was employed for improved accuracy of visual event detection, as it was relatively easy to subjectively consistently identify using video capture.

Post-processing and analysis of sEMG data was performed in Visual3D software (C-Motion Inc., Germantown, MD, USA) version 2020.07.4. A removed mean was performed on the raw EMG signals to account for the signal offset from baseline. A 40Hz high pass filter was applied to reduce movement artifacts and to attenuate low-frequency noise from the sEMG signal [36]. The sEMG signal was then full wave rectified, followed by filtering with a 15 Hz low pass filter [27-28,32,36]. Muscle activity onset and offset events were then marked for further analysis using the rectified data, resulting in 30 muscle activations isolated for the LD and RA muscles for each trial for the 15 strides identified from the synchronized video. Since the subjective toe-off events corresponded with a peak of activity of the LD and RA muscles, the entire muscle activation proceeding the toe-off event was included in the analysis, which consistently followed the left hind heel-strike events according to the validation trial. Entire muscle activation bursts proceeding the toe-off event were included in analysis, to prevent partial or split muscle activations. Within each horse, the maximum observed signal from the rectified EMG signal was found across all conditions, and used as the reference voluntary contraction (RVC) [28]. The mean rectified and maximum enveloped data for each condition were then normalized to this value. The reference voluntary contraction was measured across all conditions (with and without poles, with and without resistance bands), which was used as the reference voluntary contraction (RVC) [28]. The RVC was obtained for each subject separately, and was not the same condition across all horses. Normalization was performed separately for the walk and trot. From the normalized data, the normalized mean average rectified value (ARV), and mean maximum low pass signal were calculated over each stride for each muscle and for each individual, which were compiled for further statistical analysis.

Statistical Analysis

Statistical analysis was performed with SPSS 27 Software (IBM SPSS Statistics, IBM Inc., Armonk, NY, USA). A 2x2 repeated measures ANOVA was performed for each muscle to test for significant differences in the means of the normalized average rectified and maximum low pass values amongst the 6 subjects across the two factors; with and without resistance bands, and with and without ground poles. We tested for an interaction between the two conditions. Since no significant interactions existed between poles and resistance bands, the results of each condition could be evaluated independently. Descriptive statistics were performed (mean, standard deviation) in addition to within subject effects (p-value, partial eta squared). Values of $p \leq 0.05$ were considered significant. Where significant main effects were seen, post-hoc pairwise comparisons were performed to evaluate the differences between the conditions.

Results

Poles vs. No Poles, At A Walk (Tables 1 & 2)

The results of the electromyographic data from walking over poles are represented in Table 1 (mean, SD, p-value, and effect size) and Table 2 (post-hoc testing, poles vs. no poles). With the repeated measures ANOVA analysis, there was a significant increase in the maximum low pass values with the presence of poles in the left LD ($p = 0.045$), right LD ($p = 0.005$), left RA ($p = 0.019$), and right RA ($p = 0.015$) muscles at a walk. The percentage increase of the maximum low pass value in the left and right LD muscles in the group over poles was 51.3% and 38.0% respectively, compared to the group without poles. A relatively large increase in the maximum low pass value was seen in the left and right RA muscles of over double the maximum low pass value in the group with poles compared to without poles, 120.8% and 147.2% greater, respectively.

The presence of poles additionally resulted in a significant increase in the ARV in the right LD ($p = 0.008$), left RA ($p = 0.004$), and right RA ($p = 0.001$) muscles at a walk. The percentage increase in the ARV for the left and right RA muscles with the use of poles was 100.0% and 70.6% greater respectively compared to without poles. While the percentage increase in the mean ARV for the left and right LD muscles with the use of poles was respectively 13.5% and 27.0% greater than without poles, the increase seen in the left LD muscle did not reach statistical significance ($p = 0.087$).

Resistance Bands vs. No Resistance Bands, At A Walk (Tables 1 & 3)

The results of the electromyographic data from walking with elastic resistance bands are represented in Table 1 (mean, SD, p-value, and effect size) and Table 3 (post-hoc testing, resistance bands vs. no resistance bands). The use of resistance bands in the RA resulted in a consistent increase in the maximum low pass values and ARV, however these changes did not reach statistical significance. The maximum low pass values increased by 47.3% (p =

0.103) and 25% (p = 0.541) for the left and right RA muscles respectively. The ARV increased by 25.0% (p = 0.336) and 11.8% (p = 0.576) in the left and right RA muscles respectively when compared to without resistance bands. Additionally, the use of resistance bands in the LD muscles at the walk resulted in a non-significant reduction in both the maximum low pass value and ARV. The maximum low pass values decreased by 12.0% (p = 0.212) and 14.0% (p = 0.060) for the left and right LD muscles respectively. The ARV values decreased by 10.8% (p = 0.138) and 12.5% (p = 0.058) for the left and right LD muscles respectively when compared to without resistance bands.

Poles vs. No Poles, At A Trot (Tables 4 & 5)

The results of the electromyographic data from trotting with poles are represented in Table 4 (mean, SD, p-value, and effect size) and Table 5 (post-hoc testing, poles vs. no poles). The presence of poles at a trot resulted in a significant increase of the mean maximum low pass value of the left RA (p = 0.013) and right RA (p = 0.013) muscles. The mean maximum low pass value for the left and right RA muscles over poles was 50.9% and 60.0% greater, respectively, than the mean maximum low pass value without poles. Additionally, when comparing the ARV of the left and right RA muscles, the use of poles resulted in a 59.5% (p = 0.060) and 72.7% (p = 0.072) respective increase in ARV compared to without poles, although this was without statistical significance. The use of poles also resulted in a mean increase in the ARV of the left and right LD muscles of 20.0% (p = 0.142) and 29.8% (p = 0.094) respectively compared to the mean ARV value without poles, although this difference was not statistically significant.

Resistance Bands vs. No Resistance Bands, At A Trot (Tables 4 & 6)

The results of the electromyographic data from the trot are represented in Table 4 (mean, SD, p-value, and effect size) and Table 6 (post-hoc testing, resistance bands vs. no resistance bands). The use of elastic resistance bands resulted in a significant increase of the mean ARV of the left RA muscle (p = 0.022). For the left RA muscle, the percentage increase in the ARV in the group with resistance bands was 27.0% greater than the group without resistance bands. The right RA muscle showed a similar pattern, with a 27.2% increase in the mean ARV signal in the group with resistance bands compared to without (p = 0.619), however this did not reach statistical significance for this group. For the left and right LD muscles, the use of the resistance bands resulted in a non-significant reduction in the mean ARV value of 21.4% (p = 0.170) and 14.9% (p = 0.200) respectively compared to the group without resistance bands. Similarly, the maximum amplitude in the group with resistance bands showed a non-significant reduction in the peak low pass signal by 20.5% (p = 0.167) and 29.8% (p = 0.264) in the left and right LD muscles respectively, compared to the group without resistance bands.

Table 1: The normalized mean, standard deviation (SD), p-value, and partial eta squared (η_p^2) value of sEMG average rectified values (ARV) and maximum low pass signals in 6 horses across 15 strides at a walk, with and without ground poles and resistance bands. (LD = longissimus dorsi, RA = rectus abdominus).

	No Poles		**Poles**		**Poles Vs. No Poles**	**Resistance bands Vs. No Resistance bands**
	No Resistance Bands	**Resistance Bands**	**No Resistance Bands**	**Resistance Bands**		
ARV	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**p-vales effect size (η_p^2)**	**p-vales effect size (η_p^2)**
Left LD	0.074 (0.029)	0.066 (0.024)	0.084 (0.031)	0.072 (0.031)	p = 0.087 (0.474)	p = 0.138 (0.384)
Right LD *	0.048 (0.022)	0.042 (0.019)	0.061 (0.024)	0.050 (0.020)	p = 0.008 (0.785)	p = 0.058 (0.545)
Left RA *	0.012 (0.009)	0.015 (0.009)	0.024 (0.009)	0.025 (0.010)	p = 0.004 (0.838)	p = 0.336 (0.184)
Right RA *	0.017 (0.009)	0.019 (0.009)	0.029 (0.010)	0.030 (0.012)	p = 0.001 (0.911)	p = 0.576 (0.067)
Maximum Low pass	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**p-vales effect size (η_p^2)**	**p-vales effect size (η_p^2)**
Left LD *	0.150 (0.072)	0.132 (0.047)	0.227 (0.151)	0.153 (0.067)	p = 0.045 (0.587)	p = 0.212 (0.291)
Right LD *	0.071 (0.027)	0.061 (0.027)	0.098 (0.036)	0.086 (0.036)	p = 0.005 (0.820)	p = 0.060 (0.542)
Left RA *	0.091 (0.080)	0.134 (0.100)	0.201 (0.094)	0.222 (0.102)	p = 0.019 (0.698)	p = 0.103 (0.442)
Right RA *	0.072 (0.042)	0.090 (0.051)	0.178 (0.081)	0.185 (0.112)	p = 0.015 (0.723)	p = 0.541 (0.079)

Significant difference between poles vs. no poles (*).

Table 2: Post-Hoc pairwise comparison of of sEMG average rectified values (ARV) and maximum low pass signals of 6 horses at a walk with poles vs. no poles. (LD = longissimus dorsi, RA = rectus abdominus).

	Mean Difference (SE)	**p-value**	**95% CI Interval for the mean difference**
ARV			
Left LD	0.008 (0.004)	0.087	-0.002 to 0.018
Right LD *	0.011 (0.002)	0.008	0.004 to 0.017
Left RA *	0.011 (0.002)	0.004	0.006 to 0.017
Right RA *	0.011 (0.002)	0.001	0.007 to 0.016
Maximum Low pass			
Left LD *	0.049 (0.018)	0.045	0.002 to 0.096
Right LD *	0.026 (0.005)	0.005	0.012 to 0.040
Left RA *	0.099 (0.029)	0.019	0.024 to 0.174
Right RA *	0.100 (0.028)	0.015	0.029 to 0.172

Significant at $p < 0.05$ (*).

Table 3: Post-Hoc pairwise comparison of of sEMG average rectified values (ARV) and maximum low pass signals of 6 horses at a walk with resistance bands vs. no resistance bands. (LD = longissimus dorsi, RA = rectus abdominus).

	Mean Difference (SE)	**p-value**	**95% CI Interval for the mean difference**
ARV			
Left LD	-0.010 (0.006)	0.138	-0.025 to 0.005
Right LD	-0.009 (0.003)	0.058	-0.018 to 0.000
Left RA	0.003 (0.002)	0.336	-0.004 to 0.009
Right RA	0.001 (0.002)	0.576	-0.005 to 0.007
Maximum Low pass			
Left LD	-0.046 (0.032)	0.212	-0.128 to 0.036
Right LD	-0.011 (0.005)	0.059	-0.023 to 0.001
Left RA	0.032 (0.016)	0.103	-0.009 to 0.072
Right RA	0.013 (0.019)	0.541	-0.037 to 0.062

Significant at $p < 0.05$ (*).

Table 4: The normalized mean, standard deviation (SD), p-value, and partial eta squared (η_p^2) value of sEMG average rectified values (ARV) and maximum low pass signals in 6 horses across 15 strides at a trot, with and without ground poles and resistance bands. (LD = longissimus dorsi, RA = rectus abdominus).

	No Poles		**Poles**		**Poles Vs. No Poles**	**Resistance bands Vs. No Resistance bands**
	No Resistance Bands	**Resistance Bands**	**No Resistance Bands**	**Resistance Bands**		
ARV	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**p-vales effect size (η_p^2)**	**p-vales effect size (η_p^2)**
Left LD	0.070 (0.028)	0.055 (0.033)	0.084 (0.051)	0.064 (0.042)	p = 0.142 (0.377)	p = 0.170 (0.339)
Right LD	0.047 (0.023)	0.040 (0.024)	0.061 (0.032)	0.059 (0.028)	p = 0.094 (0.460)	p = 0.200 (0.304)
Left RA †	0.037 (0.016)	0.047 (0.013)	0.059 (0.018)	0.065 (0.024)	p = 0.60 (0.538)	p = 0.022 (0.681)
Right RA	0.033 (0.011)	0.042 (0.010)	0.057 (0.021)	0.053 (0.014)	p = 0.72 (0.507)	p = 0.619 (0.053)
Maximum Low pass	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**Mean (SD)**	**p-vales effect size (η_p^2)**	**p-vales effect size (η_p^2)**
Left LD	0.297 (0.096)	0.236 (0.134)	0.312 (0.203)	0.233 (0.154)	p = 0.853 (0.008)	p = 0.167 (0.343)
Right LD	0.275 (0.197)	0.193 (0.159)	0.241 (0.193)	0.227 (0.154)	p = 0.997 (0.000)	p = 0.264 (0.240)
Left RA *	0.210 (0.113)	0.241 (0.069)	0.317 (0.111)	0.343 (0.089)	p = 0.013 (0.742)	p = 0.114 (0.423)
Right RA *	0.195 (0.078)	0.256 (0.063)	0.312 (0.124)	0.296 (0.083)	p = 0.013 (0.741)	p = 0.238 (0.264)

Significant difference between poles vs. no poles (*). Significant difference between resistance bands and no resistance bands (†).

Table 5: Post-Hoc pairwise comparison of of sEMG average rectified values (ARV) and maximum low pass signals of 6 horses at a trot with poles vs. no poles. (LD = longissimus dorsi, RA = rectus abdominus).

	Mean Difference (SE)	**p-value**	**95% CI Interval for the mean difference**
ARV			
Left LD	0.012 (0.007)	0.142	-0.006 to 0.029
Right LD	0.016 (0.008)	0.094	-0.004 to 0.037
Left RA	0.020 (0.008)	0.060	-0.001 to 0.041
Right RA	0.018 (0.008)	0.072	-0.002 to 0.038
Maximum Low pass			
Left LD	0.006 (0.032)	0.853	-0.075 to 0.088
Right LD	0.000 (0.039)	0.997	-0.100 to 0.100
Left RA *	0.105 (0.028)	0.013	0.034 to 0.176
Right RA *	0.079 (0.021)	0.013	0.025 to 0.132

Significant at $p < 0.05$ (*).

Table 6: Post-Hoc pairwise comparison of of sEMG average rectified values (ARV) and maximum low pass signals of 6 horses at a trot with resistance bands vs. no resistance bands.

	Mean Difference (SE)	**p-value**	**95% CI Interval for the mean difference**
ARV			
Left LD	-0.018 (0.011)	0.170	-0.046 to 0.011
Right LD	-0.005 (0.003)	0.200	-0.013 to 0.004
Left RA †	0.008 (0.0032)	0.022	0.002 to 0.015
Right RA	0.002 (0.004)	0.619	-0.009 to 0.014
Maximum Low pass			
Left LD	-0.070 (0.043)	0.167	-0.182 to 0.042
Right LD	-0.048 (0.038)	0.264	-0.146 to 0.050
Left RA	0.029 (0.015)	0.114	-0.010 to 0.068
Right RA	0.022 (0.017)	0.238	-0.021 to 0.065

Significant difference between resistance bands and no resistance bands (†).

Discussion

Surface EMG is a useful technique for investigating function of the neuromuscular system in horses [27-28,31-32]. To characterize changes in the neuromuscular response with therapeutic interventions, this study reported two variables across filtered sEMG data including the average rectified value (ARV) and the maximum value from the low pass signal. The ARV is a measure suggesting average "work done" over a specified time interval and was calculated as the mean value of the full wave rectified sEMG signal over one complete gait cycle [27-30]. The maximum low pass value represents peak activity, which was also examined as some therapeutic interventions may result in brief contractions with a high peak of activity, versus longer sustained contractions with a low peak of activity [34]. To document which changes are occurring with the use of ground poles or resistance bands, both values were reported for the muscles investigated. To facilitate comparative analysis of muscle activity using sEMG, a maximal reference voluntary contraction (RVC) is employed for normalization. In humans, a maximum voluntary contraction (MVC) is used as a 100% effort measure, which can be compared to subsequent muscle activations across conditions or activities for that individual. However, it is recommended to use a RVC as a pseudoreference state for comparative analysis in cases when a MVCs is difficult or impossible to obtain, such as in animals or in humans with pain or neurologic disorders [27-28].

In this study, the LD muscle was examined given its role as one of the major spinal stabilizing muscles in horses with reported functions of spinal extension and providing lateral stability [6,13-17]. When walking over ground poles compared to without, there was a significant increase in the maximum low pass value in both left and right LD muscles correlating to larger peaks of activation. These increased peaks of LD activity may be correlated with stabilization of the axial skeleton in response to perturbations of the horse's balance as the swing limbs are raised over the ground pole obstacles [10]. With no change in withers or croup height identified with ground poles, the LD muscle is not needed for an increased suspension phase as seen with jumping, but instead may be stabilizing the axial skeleton given the increased flexions of forelimbs and hindlimbs and resulting rotation of the trunk [5,25-26]. This increase in maximum low pass value was not seen with ground poles at a trot. Previous reports have documented a linear increase in LD activity with speed, to oppose inertial forces of the visceral mass which increase with speed [13]. Therefore, it is possible that the absence of a significant increase in peak low pass sEMG activity of the LD at a trot over poles may have been related to the relative increase in baseline peak sEMG activity at a trot without poles.

There was additionally an increase in the ARV of both the left and right LD muscles by 13.5% and 27% respectively when walking over poles compared to without poles, suggesting a greater average work done across each stride, however this was only statistically significant in the right LD muscle. Previous

sEMG studies have documented bilateral contraction of the longissimus dorsi muscle during thoracolumbar extension when traveling in a straight line, and with increased sEMG intensity of the inside LD muscle when traveling in a circle or lateral bending [13-16]. Lateral biases have been documented in the LD during static flexion and extension, which may be due to multiple factors including individual horse's laterality preference or handedness [13,15]. While training factors have also been postulated to be a factor for differences in muscle activation on either side of the body, this is less likely the case in the included population of untrained pasture-rested horses [13]. At a trot there was also an increase in the mean ARV in both the left and right LD muscles by 20.0% and 29.8% respectively with ground poles compared to without poles, however this did not reach statistical significance in either muscle. Further research combining surface electromyographic activity with motion capture or inertial measurement unit (IMU) sensors throughout data collection would enable characterization of the timing of induced changes in muscle activation with poles relative to phases in the gait cycle. For example, with an increase in the sEMG activity with ground poles, it could be further documented if this increase in muscle activation is occurring during forelimb and hindlimb protraction to offset axial rotation of the spine, or possibly during late swing phase for spinal stabilization during hindlimb retraction.

The core abdominal musculature is also important for providing spinal stiffness, maintaining balance and posture, and for controlling the body's angular momentum when maneuvering over obstacles [9-10,13,40]. Increased maximal activation of the rectus abdominus has been previously reported in three of six horses trotting over ground and raised poles compared to working on the flat, while mean workload was reportedly unchanged [11]. In this study both the left and right RA muscles had a significant increase in maximum low pass value of the sEMG signal with the presence of ground poles in both the trot and walk. This suggests that at both gaits, the RA bilaterally is having larger peaks of electrical activity, corresponding with a greater core abdominal muscular response with the presence of ground poles. Additionally, at a walk, the right and left RA had a significant increase in the ARV with the presence of ground poles, which suggests greater average work done across each stride. However, an increase in the ARV was not seen at a trot over ground poles, which fits with previously reported data where no change in mean workload was found [11].

The use of elastic resistance bands as a proprioceptive training tool were also of interest in this study. They are used as a training aid to improve horse posture and core strength for injury prevention and for improving performance of sport horses [1,3,6,41]. Engagement of hindquarter and abdominal musculature is thought to improve dynamic stability of the back and pelvis through core postural muscle development [22,26,36]. After a 4-week training program using an elastic resistance band system, several significant kinematic effects were found. This included a reduction in withers roll, withers pitch, and a reduction in thoracolumbar mediolateral movement, consistent with improved dynamic

stability of the vertebral column [3]. To further investigate the kinematic changes induced by the elastic resistance bands, this study aimed to characterize baseline sEMG activity of selected epaxial and abdominal muscles. At a trot, both the left and right RA muscles demonstrated a respective 27% and 27.2% increase in the mean ARV signal in the group with resistance bands compared to without, however this only reached statistical significance in the left RA. This indicates a significant increase in the average work done by the left RA across each stride at a trot with the presence of the elastic resistance bands. At a walk with elastic resistance bands, the observed increase in the ARV signal and maximum low pass values of both the left and right RA muscles compared to without elastic resistance bands did not reach statistical significance. It is possible that with a larger sample size, a significant change in RA activity may have been obtained in the right RA at a trot. As discussed with the LD muscle over ground poles, lateral biases have been identified in certain paired muscle groups and can be due to factors such as training, acquired pathologies, handedness or laterality preference, or due to specific muscle function [27].

The other observation with the use of elastic resistance bands was a reduction in the mean ARV of the left and right LD muscles at a walk and trot, as well as a reduction in the maximum low pass value of the left and right LD muscles at a walk. Further research is needed to document the possible statistical significance of this reduced LD activation with elastic resistance bands. The activity of the LD muscles have been investigated with the use of side reins and the Pessoa system, which are other training aids used in horses that aim to develop the epaxial muscles and increase stabilization of the back [6,38]. The sEMG intensity of the LD muscles were not increased by the training aids but were the greatest for the control conditions at the walk and trot [6]. Theories for altered LD activity with the use of training aids include a reduction in spinal extension, increased spinal stability with the presence of the training aid, or increased eccentric versus concentric activity of the LD [25,27]. Further investigation is needed to better characterize these observed changes in LD activity.

One of the limitations of this study was that the electromyographic activity could not be directly or objectively correlated to distinct phases of the gait cycle, given the absence of objectively defined gait events. The use of a 3D motion capture system in this study was limited to checking the firing sequences of the observed muscles with known gait events to provide a confirmation of the phasic activation patterns across the gait cycle. This validation of the activation pattern of the LD and RA muscles within the gait cycle increased confidence in the subjective stride selection technique based on toe-off events from video capture and the subsequent marking of muscle onsets and offsets within the isolated strides. Motion capture was not available for use during data collection with application of elastic resistance bands and over poles. While sEMG activity was able to be compared across activations within a consistent number of strides, the absence of motion capture or inertial measurement unit sensors (IMUs)

throughout the data collection limited more detailed analysis regarding the timing of muscle activations during the gait cycle.

Another limitation for the current investigation was the small population of available horses. Available horses were also of different breeds and signalments, leading to a small diverse sample population. While two horses used in this study were crossed with gaited horse breeds (Tennessee Walking Horse, Saddlebred), both individuals performed a diagonal two-beat trot for data collection, and any strides of irregular gaiting identified live or on video capture were excluded from the analysis. Utilizing horses of different breeds and body types may increase inter-subject variance due to factors such as differences in muscle fiber, skin thickness, depth of subcutaneous fat, or distribution of sweat glands and sweat production [27]. Additionally, this study involved a healthy population of horses, and these findings cannot be extrapolated to horses with pathology, such as back pain or lameness. While horses were screened for lameness at the onset of the study, all sample subjects were administered a low-dose non-steroidal anti-inflammatory drug (phenylbutazone, 2.2mg/kg, once daily) throughout the data collection period to prevent exercise-induced muscle soreness, given that subjects were not maintained in routine exercise. Fitness, level of training, and fatigue are important variables to consider as they can influence adaptations in muscle tissue (fiber hypertrophy, remodeling of fiber type, improvements in timing or firing synchronization, etc.) which can influence the resulting sEMG signal [27]. All subjects were of a similar baseline level of fitness, as they were all housed in pasture with no forced or routine exercise.

For consistency of data collection, exercise protocols were completed in the same order, and with an hour break between conditions without resistance bands and with resistance bands. Given the relatively light work at a walk and trot and the hour break half way through data collection, it is unlikely that muscle fatigue was a relevant factor during data collection for this study, and therefore randomization of the study conditions should have minimally affected the sEMG results. Additionally, it has been documented that increasing speed of gait can increase muscle recruitment in the horse [13,27]. Some studies measure and account for this with use of an accelerometer. Although we appreciate that using speed as a covariate may allow for some variability, we did not record or regulate speed, and used similar methods to other studies using the horse's preferred speed [28,42]. An additional factor involved with the analysis of the therapeutic exercise conditions was testing both elastic resistance bands and ground poles simultaneously to investigate if both training tools had an additive effect in influencing muscle activation. With use of the 2x2 repeated measures ANOVA, we tested for interactions between ground poles and elastic resistance bands. However, no significant interactions were identified, therefore it can be concluded that the effects of one therapeutic strategy did not influence the effects of the other in this study.

For future study, repeating sEMG measurements over ground poles and with elastic resistance bands with a motion capture system would enable

investigation of the timing of the muscle activation relative to known gait events within the gait cycle, in addition to better characterizing changes in muscle recruitment patterns, or types of activity such as concentric versus eccentric types of muscle contractions. Additionally, the combined use of a motion capture system would also allow for evaluation of kinematic changes occurring with various therapeutic conditions applied, such as changes in stride length, posture, or joint characteristics. Some other considerations include the amount of tension applied to the elastic resistance bands, as the physiological influence may change according to alterations in tension, or stored and released energy in the elastic resistance bands. While 25% stretch was used in this study, earlier research investigated elastic resistance bands under 30% stretch, and up to 50% stretch is recommended by manufacturers of an elastic resistance band system (Equicore Concepts LLC, East Lansing, Michigan, USA) [3]. Another consideration is the incorporation of a training period as the horse's posture, spinal stability, muscle activation patterns, and possible "learned" neuromuscular responses may alter with training. Clinically, it would be interesting to evaluate horses with the elastic resistance bands applied with different amounts of tension, and to watch for kinematic and electromyographic changes across a several week training period. Additionally, looking at horses of similar breed, discipline, or training level would lead to a more uniform sample population. After further documentation of the effects of certain therapeutic strategies in normal horses, this information could be applied to horses with lameness or certain pathologies, such as back pain, lumbosacral or sacroiliac restriction, or disuse atrophy.

In conclusion, the use of ground poles resulted in a significant increase in sEMG activity in the RA muscles bilaterally at a walk and trot. Additionally, a significant increase in the sEMG activity of the LD muscles bilaterally was seen with ground poles at the walk. This shows that the incorporation of ground poles in a rehabilitation program is expected to increase core abdominal activation and epaxial muscle recruitment. The use of elastic resistance bands at 25% stretch demonstrated a significant increase in sEMG activity in the left RA muscle at the trot. Results from this study require further investigation to better understand the effects of elastic resistance bands on muscle activation, particularly with reference to the type of muscle activation, the stages of the gait cycle, at different degrees of stretch, and the effect of a training period.

CHAPTER II
Equine Kinematics

A version of this chapter has not been previously published.

This work was completed with the guidance of mentor and collaborator Dr. Songing Zhang, and data collection and processing was performed in collaboration with Dr. Becca Noel.

Abstract

Influencing a horse's posture to achieve "rounding" or flexing of the spine is a common goal of many riding and training strategies. This study aimed to quantify changes in thoracolumbar, lumbosacral, and tarsus joint angles in a horse traveling in a straight line at a walk and trot with the presence of elastic resistance bands, and with the presence of a "passive" rider, or "active" rider using leg, seat, and rein aids to encourage increased spinal flexion and engagement of the hindquarters. Abdominal elastic resistance bands applied at 25% stretch. Reflective markers were secured to pre-determined landmarks on the horse's body, which were used for tracking the horse through space with a 12-camera 3D motion capture system. After data collection at a walk and trot with conditions with elastic resistance bands, and with a "passive" and "active" rider, the kinematic marker data were processed using Visual3D software so that joint angle calculations could be made. For the joints of interest, the joint angles were tracked during the stance and swing phases. Additionally, the maximum joint angles, minimum joint angles, joint range of motion, and metric means were calculated for the thoracolumbar, lumbosacral and tarsus joints during stance phase. According to preliminary data, the presence of a "passive" rider resulted in increased thoracolumbar extension at a walk, while the presence of an "active" rider resulted in increased thoracolumbar and lumbosacral flexion at a walk and trot. The "passive" and "active" rider resulted in increased tarsus extension at a walk and trot. The presence of elastic resistance bands resulted in an increase of flexion of the thoracolumbar region at a walk and trot with and without a rider, increased lumbosacral flexion with the presence of an "active" rider, and increased tarsus extension at both a walk and trot without a rider.

Aim

The main purpose of this pilot investigation was to assess the change in kinematics of the tarsus, thoracolumbar and lumbosacral regions when using elastic resistance bands with and without a rider in walk and trot. Additionally, we aimed to evaluate; (1) the effect of an "active" rider that is utilizing leg, seat, and rein aids to influence horse posture and impulsion, as compared to a "passive" rider on a loose rein with no additional aids, and (2) if the elastic resistance bands and rider have an additive or synergistic effect on induced postural changes in the horse.

We hypothesized that the use of elastic resistance bands would result in an increase in thoracolumbar and lumbosacral flexion, and an increase in hock range of motion. We also hypothesize that an “active” rider will cause a further increase in these changes.

Materials And Methods

Marker Set

To facilitate collection of kinematic data, a marker set was created so that segment and joint characteristics could be investigated along the horse’s spine, forelimb and hindlimb on the left side of the body (*see Appendix I: Marker Set)*. The axial skeleton consisted of 6 segments; head, cranial cervical, caudal cervical, thoracic, lumbar, and sacral. The left forelimb was divided into five segments; shoulder, humerus, radius, metacarpus, and pastern-foot. The left hindlimb was also divided into five segments; pelvis, femur, tibia, metatarsus, and pastern-foot. To track each segment, anatomic markers were placed over palpable anatomical landmarks which were used for the static calibration trial, and tracking markers were placed over the body segments which were used during dynamic trials. Each segment, with an exception of the head segment, consisted of a minimum of two anatomic markers and four tracking markers. The four tracking markers for each segment were made into clusters by mounting them onto a rigid thermoplastic sheet in a non-collinear orientation. The thermoplastic sheets were first heated and molded to the horse at the various segment locations, to maximize contour for a more secure and stable fit. The head segment did not have a cluster but instead consisted of four anatomic markers which remained in place for the dynamic trials to be used as tracking markers. One virtual anatomic marker was created at the location of the sacrum joint, which was digitally calculated with Visual3D software (C-Motion Inc., Germantown, MD, USA) version 2020.07.4, at 50% the distance between the left and right mid-tuber coxae markers. For this reason, the bilateral mid-tuber coxae markers were left in place for the dynamic trials as tracking markers.

Subject

One horse was used for this pilot investigation, who was a 16-year old quarter horse mare, and had been previously trained and used as a show horse. The protocol used for data collection has been accepted by the University of Tennessee Institutional Animal Care and Use Committee (IACUC).

Instrumentation

The horse was instrumented with 14mm diameter reflective pearl markers (B&L Engineering, Division of Pinsco, Inc., Santa Ana, CA, USA) which were applied at pre-determined locations using palpable surface landmarks (Appendix I), and secured with double sided adhesive strips and methyl methacrylate adhesive glue (Figure 4). Cluster marker sets consisting of four 14mm pearl markers affixed to pre-molded thermoplastic plates were secured by Velcro strips

(Velcro USA Inc., Manchester, NH, USA) to customized neoprene leg wraps on the distal limbs, and to adhesive foam sheets for the body or proximal limbs (Reston, 3M, St. Paul, MN, USA) (Figure 4). These marker clusters were used for tracking purpose. All neoprene wraps were secured by adhesive tape, and all adhesive sheets were secured by methyl methacrylate adhesive glue.

The horse was placed in an English saddle using a saddle pad manufactured with buckles for the application of hindquarter and abdominal elastic resistance bands (Equicore Concepts LLC, East Lansing, Michigan, USA). The elastic resistance bands were applied behind the hindquarter musculature and ventral abdomen with a 25% stretch, which was calculated from 75% of the distance between attachment points of the elastic bands. For the ridden conditions, a bridle was used which the horse had previously been fit and accustomed to (Figure 5). For non-ridden conditions, the horse was handled in a leather halter.

Data Collection

Kinematic data was collected using a 12-camera 3D motion capture system (VICON Motion Systems Ltd, Oxford, UK) at a sampling frequency of 100 frames/second (Hz). Following instrumentation with both anatomic and tracking markers, the horse was held still in the capture area for a static calibration trial. Subsequently, the anatomic markers were removed before the commencement of the dynamic trials (with the exception of the markers comprising the head segment and the bilateral mid-tuber coxae markers). Six conditions were performed at a walk followed by trot, in a straight line on an asphalt lameness pad in the following non-randomized order;

1. Without elastic resistance bands, no rider
2. Elastic resistance bands, no rider
3. Without elastic resistance bands, “passive” rider
4. Without elastic resistance bands, “active” rider
5. Elastic resistance bands, “passive” rider
6. Elastic resistance bands, “active” rider

The horse performed 5 separate trials for each of the conditions at the walk and trot. One rider was used for consistency between ridden conditions.

Data Processing

Marker labeling was performed in Nexus 2.10 software (VICON Motion Systems Ltd, Oxford, UK), using a VSK custom labelling skeleton. This VSK custom labeling skeleton could be applied to each individual trial for automated marker labeling, which was then manually checked for accuracy by two separate individuals. Following custom labeling, gap filling was performed using a cubic spline filling algorithm for all small gaps, and with a pattern fill algorithm for any



Figure 4. Marker set with anatomic and tracking markers in place.



Figure 5. Horse with rider, elastic resistance bands, and tracking markers (anatomic markers removed).

larger gaps. The cubic spline filling algorithm uses existing trajectory data to predict the missing values, which is relatively accurate for small gaps. The pattern fill algorithm involves selecting nearby markers (preferentially on the same cluster) that exhibit similar motion to the missing marker to generate the missing trajectory. The majority of the larger gaps were located at fore and hind pastern clusters, which enabled selection of the adjacent markers fixed to the same rigid thermoplastic cluster for gap filling by pattern fill.

Further processing and analysis of the kinematic data was performed in Visual3D 3D biomechanics analysis suite (version 2020.07.4, C-Motion Inc., Germantown, MD, USA). The marker data was filtered with a 4^{th} order Butterworth lowpass filter with zero-lag at a cutoff frequency of 8 Hz, and interpolated. With the absence of force data for this data set, gait events were defined as follows; using the peak of horizontal velocity of the center of mass of the left hind pastern-foot segment to determine heel-strike, and using the minimum in horizontal acceleration of the center of mass of the left hind pastern-foot segment to identify toe-off (Figure 6). The animation across all trials was examined and checked to confirm evaluated consistency of the gait events for both the walk and trot.

This method of identifying heel-strike and toe-off events was found consistent for both walk and trot trials by two separate observers. Two consecutive left hind heel-strike events and one left hind toe-off event were marked for each trial, to enable separate analysis of stance phase and swing phase events. For each condition, a minimum of 3 trials containing a complete gait cycle for the left hindlimb were utilized for further analysis.

The joint angles were computed using a joint coordinate system and follow the X-Y-Z rotational sequence. The joint angles are expressed using a right-hand rule and the y and z axes were negated. The proximal segment was treated as the parent segment. Joint angles were calculated for (1) thoracolumbar joint, between thoracic and lumbar segments, (2) lumbosacral joint, between lumbar and sacral segments, and (3) left tarsus, between tibia and metatarsal segments. The relationship between these segments in the static skeleton can be seen in Figure 7. Discrete peak values including maximum angle, minimum angle, joint range of motion (maximum angle minus minimum angle), and mean were identified or computed in Visual3D.

Results

When evaluating the results of this pilot data, statistical analyses were not performed as there was only one sample subject. The angle of each joint of interest was graphically represented for both the stance and swing phase with a solid line for the mean angle, and a shaded line representing the standard deviation. The maximum joint angle, minimum joint angle, joint range of motion,

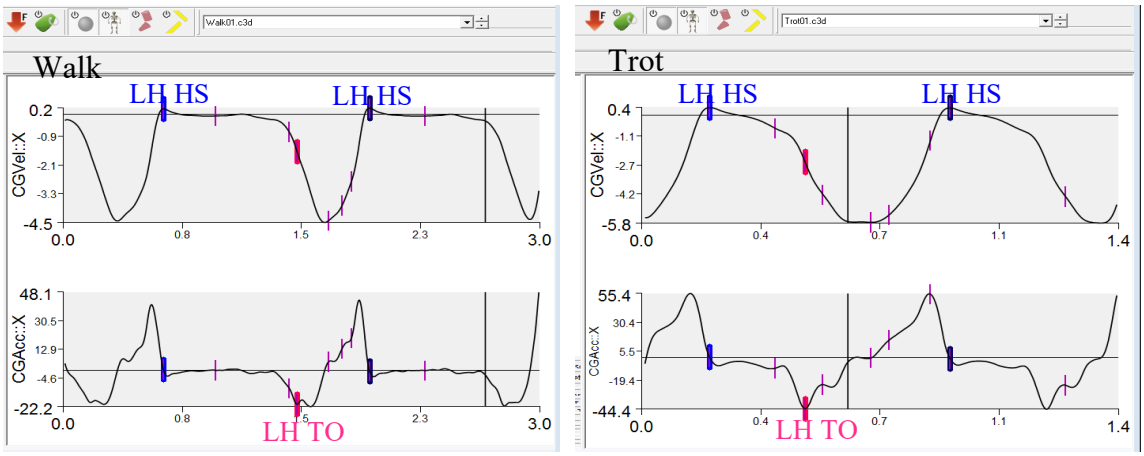


Figure 6. Defining gait events in Visual3D using peak in horizontal velocity of the center of gravity of the left hind pastern-foot segment for left hind heel-strike (LH HS), and minimum in horizontal acceleration of the center of gravity of the left hind pastern-foot segment for left hind toe-off (LH TO).

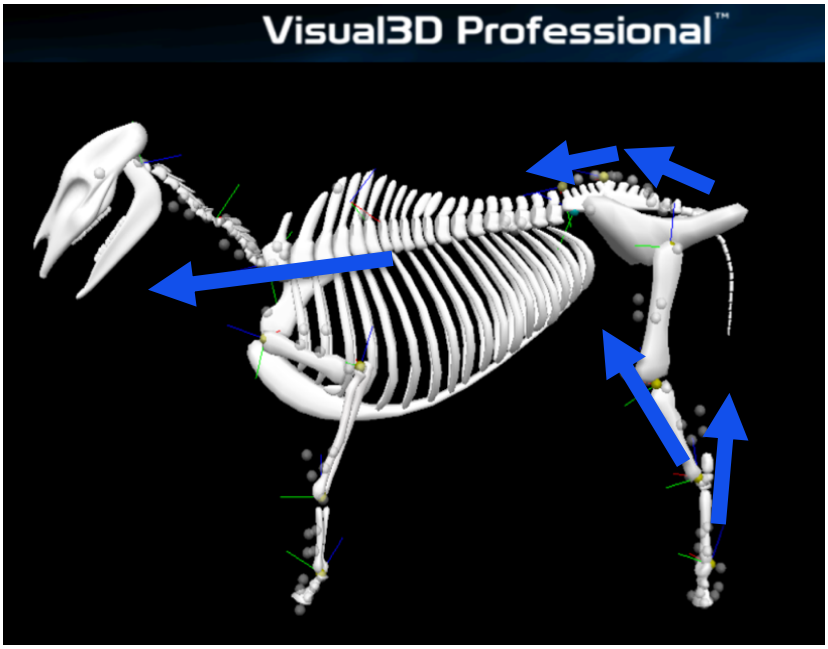


Figure 7. Relationship between segments at the regions of interest in the static skeleton.

and metric mean joint angle were calculated for each joint of interest during stance phase.

Thoracolumbar Angle, At A Walk (Figures 8 & 9, Table 7)

The thoracolumbar angle was evaluated on the sagittal plane for changes in flexion or extension of this joint under different conditions. The convention for the relationship between the thoracic and lumbar segments is that increased or positive angle values represent thoracolumbar joint extension. When evaluating the effect of a rider on the thoracolumbar joint angle (Figure 8), the greatest relative joint flexion is seen with the “active” rider, followed by no rider. Whereas the “passive” rider results in greater joint extension compared to the other two conditions. This was demonstrated graphically (Figure 8) as well as with the trend in the highest thoracolumbar maximum angle in the passive rider, and the lowest thoracolumbar minimum angle in the active rider (Table 7). This suggests that while the thoracolumbar joint is flexing and extending during stance phase, an overall trend of increased relative flexion is seen in the active rider, followed by no rider, and with reduced relative joint flexion observed with the passive rider. The overall range of motion of this joint did not appear to differ greatly between conditions.

When evaluating the effect of the elastic resistance bands (ERB) on thoracolumbar angle (Figure 9), a discrete separation in the mean angle and shaded standard deviations lines are seen, characterized by increased relative flexion with the elastic resistance bands compared to without elastic resistance bands in both “passive” and “active” rider conditions. A similar trend was seen with no rider with increased thoracolumbar flexion demonstrated with elastic resistance bands compared to without, from approximately 40% of stance phase onward.

Lumbosacral Angle, At A Walk (Figures 10 & 11, Table 8)

The lumbosacral angle was evaluated on the sagittal plane for changes in flexion or extension of this joint under different conditions. The convention for the relationship between the lumbar and sacral segments is that increased or positive angle values represent lumbosacral joint extension. When evaluating the effect of a rider on the lumbosacral joint angle (Figure 10), a trend is observed with increased overall joint flexion observed with both rider conditions over no rider. More specifically, during early stance, the active rider displayed the greatest relative joint flexion (Table 8). However, during approximately 75% stance at the period of greatest joint flexion, both the active and passive riders displayed a similar level of maximal joint flexion, which was increased by approximately 0.7 degrees compared to without a rider. Additionally, the smallest relative range of motion (or difference between maximal flexion and extension) was seen in the active rider condition, with the largest relative range of motion seen in the passive rider condition. When evaluating the effect of the

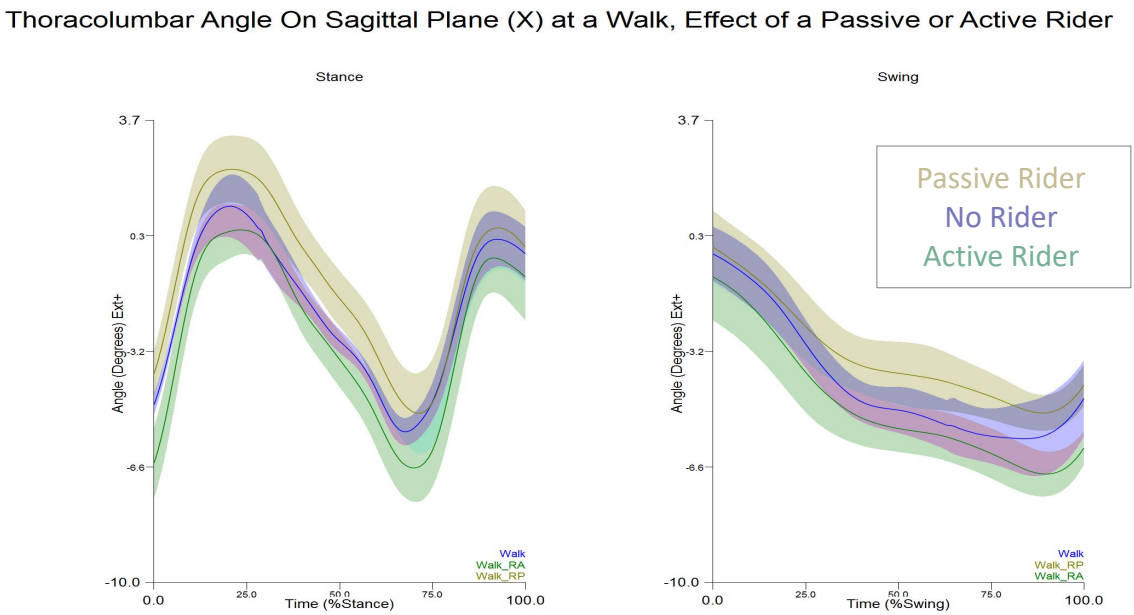


Figure 8. Effect of a "passive" or "active" rider on thoracolumbar angle at a walk.

Table 7: Thoracolumbar (TL) angle at a walk, with different rider conditions (no rider, "passive" rider, and "active rider"), and with and without elastic resistance bands (ERB).

Walk	TL Angle Max		TL Angle Min		TL ROM		TL Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	1.090	0.775	-5.935	0.874	7.025	0.397	-1.964	0.579
No Rider +ERB	0.004	1.662	-7.531	1.061	7.536	0.631	-3.593	1.348
Passive Rider	2.283	1.011	-5.391	0.948	7.973	0.238	-4.436	0.238
Passive Rider +ERB	0.112	0.654	-7.693	0.548	7.805	0.997	-3.301	0.281
Active Rider	0.490	0.739	-6.919	0.710	7.409	0.461	-2.577	0.870
Active Rider +ERB	-0.861	0.280	-8.835	0.263	7.696	0.787	-0.908	0.985

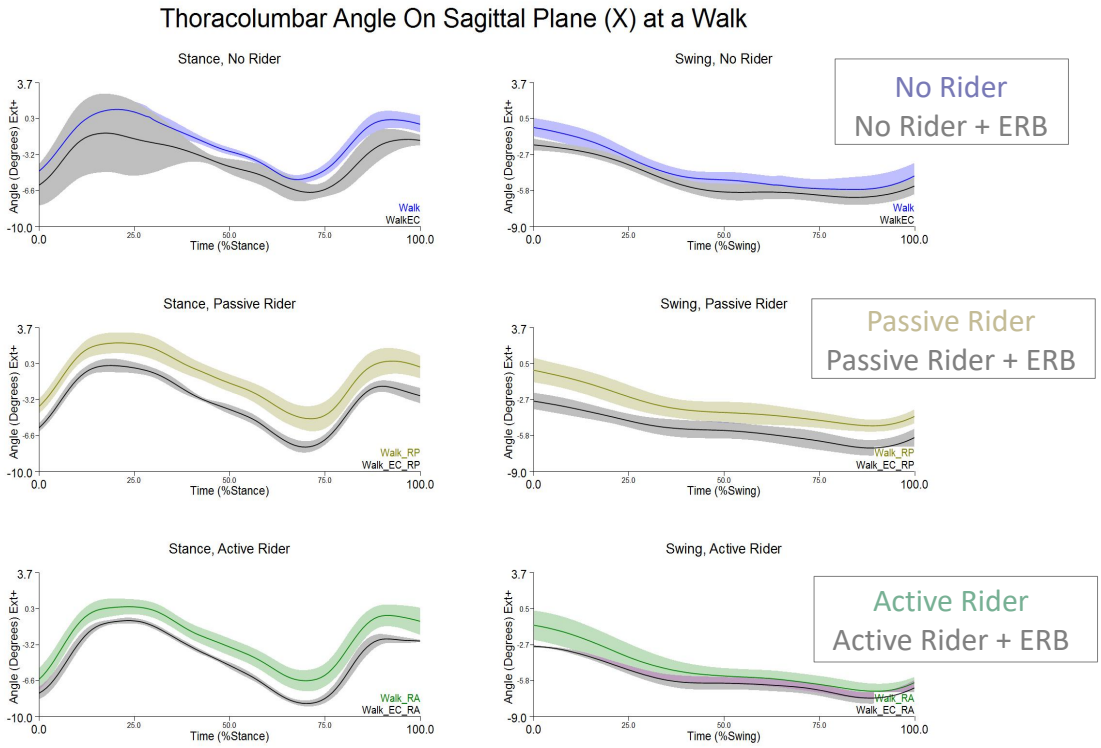


Figure 9. Effect of elastic resistance bands (ERB) on thoracolumbar angle at a walk.

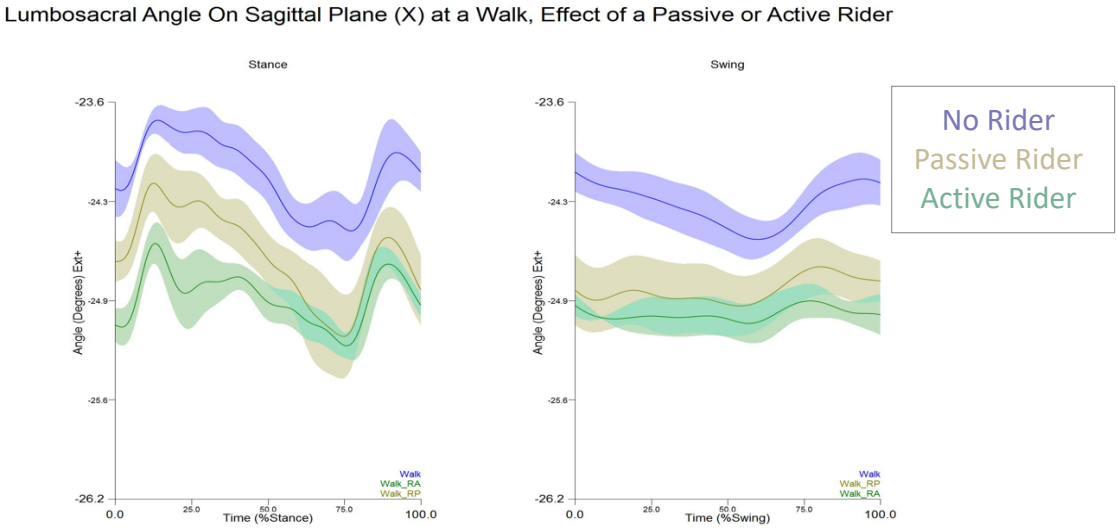


Figure 10. Effect of a “passive” or “active” rider on lumbosacral angle at a walk

elastic resistance bands on the lumbosacral joint angle (Figure 11), no discrete separation of the means or standard deviations was seen, suggesting no further appreciable difference in lumbosacral joint angles was induced with the presence of the elastic resistance bands.

Tarsus Angle, At A Walk (Figures 12 & 13, Table 9)

The tarsus angle was evaluated on the sagittal plane for changes in flexion or extension of this joint under different conditions. The convention for the relationship between the tibia and metatarsus segments is that increased or positive angle values represent tarsal joint flexion. When evaluating the effect of a rider on the tarsus joint angle (Figure 12), a trend is observed with increased overall joint extension observed with both rider conditions compared to no rider.

There is also a much narrower standard deviation in both rider conditions, while the variability in overall flexion and extension in the condition without a rider is much broader. When evaluating the effect of the elastic resistance bands (Figure 13), there was no appreciable change in tarsus flexion or extension seen in either of rider conditions, demonstrated by the superimposition of both curves graphically. However, an overall relative increase in tarsus extension was seen with the presence of elastic resistance bands with no rider.

Thoracolumbar Angle, At A Trot (Figures 14 & 15, Table 10)

When evaluating the effect of a rider on the thoracolumbar joint angle at a trot (Figure 14), there was a slight change in the timing of joint flexion and extension characterized by an earlier peak in maximal extension with no rider at approximately 50% stance, whereas the maximal extension was seen in both rider conditions at approximately 55 - 60% stance followed by a quicker transition to maximal flexion. There was also increased variability in both rider conditions demonstrated by increased standard deviation of the mean joint angles as compared to the no rider condition. This deviation in timing of thoracolumbar joint flexion and extension was also seen during the swing phase, where the condition without the rider was quicker to reach maximum extension at approximately 35 - 40% of swing relative to the conditions with the rider that reached peak extension closer to 50 – 55% of swing phase. The overall range of motion was similar between rider conditions.

When looking at the maximum degree of thoracolumbar joint flexion obtained, there was approximately 0.4 – 0.45 degrees increased relative flexion in the active and passive rider conditions respectively compared to without the rider (Table 10). Graphically there is also a clear distinction between the amount of thoracolumbar joint flexion with the presence of the elastic resistance bands (Figure 15), characterized by increased joint flexion in all three conditions without a rider and with “passive” and “active” riders. Given the separation of the mean joint angle with no superimposition of the shaded standard deviation lines, a distinct difference with the elastic resistance bands can be appreciated. This

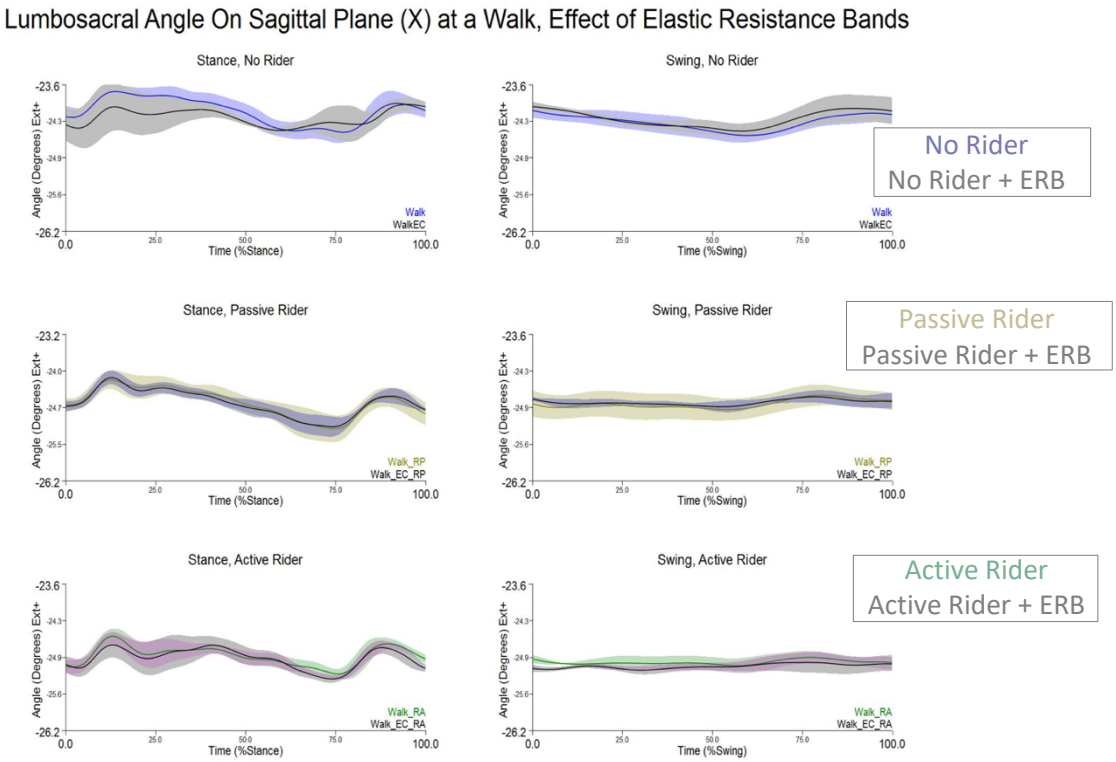


Figure 11. Effect of elastic resistance bands (ERB) on lumbosacral angle at a walk.

Table 8: Lumbosacral (LS) angle at a walk, with different rider conditions (no rider, "passive" rider, and "active rider"), and with and without elastic resistance bands (ERB).

Walk	LS Angle Max		LS Angle Min		LS ROM		LS Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	-23.702	0.101	-24.470	0.170	0.767	0.119	-24.069	0.138
No Rider +ERB	-23.790	0.057	-24.604	0.219	0.814	0.193	-24.182	0.115
Passive Rider	-24.120	0.180	-25.140	0.275	1.020	0.172	-24.589	0.197
Passive Rider +ERB	-24.080	0.143	-25.097	0.205	1.017	0.299	-24.598	0.084
Active Rider	-24.520	0.136	-25.216	0.092	0.695	0.116	-24.871	0.110
Active Rider +ERB	-24.613	0.175	-25.313	0.065	0.699	0.140	-24.918	0.135

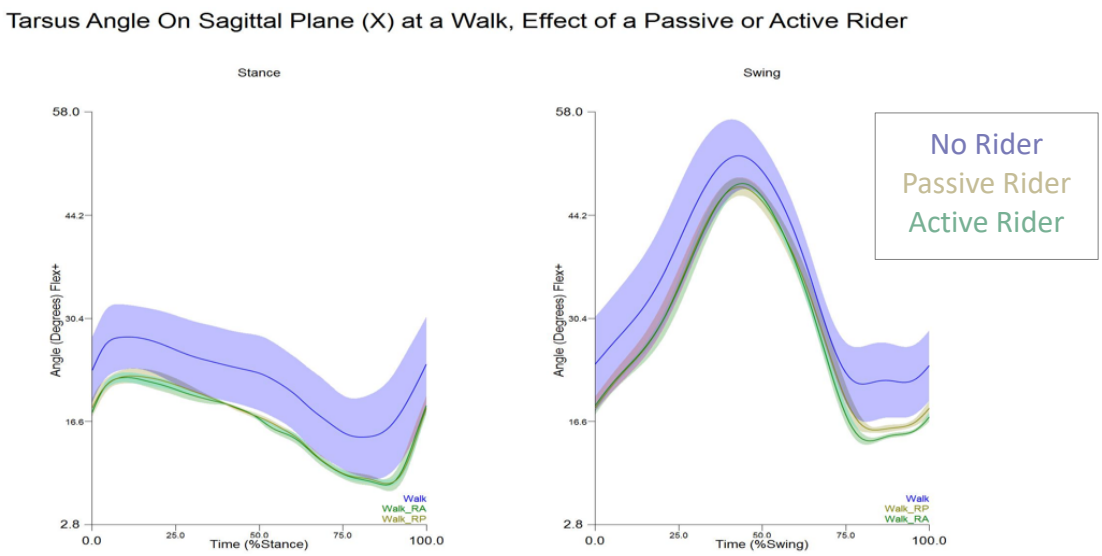


Figure 12. Effect of a "passive" or "active" rider on tarsus angle at a walk.

Table 9: Tarsus angle at a walk, with different rider conditions (no rider, "passive" rider, and "active rider"), and with and without elastic resistance bands (ERB).

Walk	Tarsus Angle Max		Tarsus Angle Min		Tarsus ROM		Tarsus Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	52.322	4.523	14.283	5.494	38.039	1.113	21.954	5.009
No Rider +ERB	46.208	1.237	5.232	0.873	40.976	1.434	14.636	0.213
Passive Rider	48.006	1.174	8.259	0.428	39.747	1.241	16.295	0.315
Passive Rider +ERB	47.991	2.000	8.340	0.362	39.651	2.301	16.176	0.356
Active Rider	48.521	0.690	8.027	0.763	40.494	0.849	16.098	0.505
Active Rider +ERB	48.015	1.414	7.683	0.517	40.332	1.726	15.924	0.274

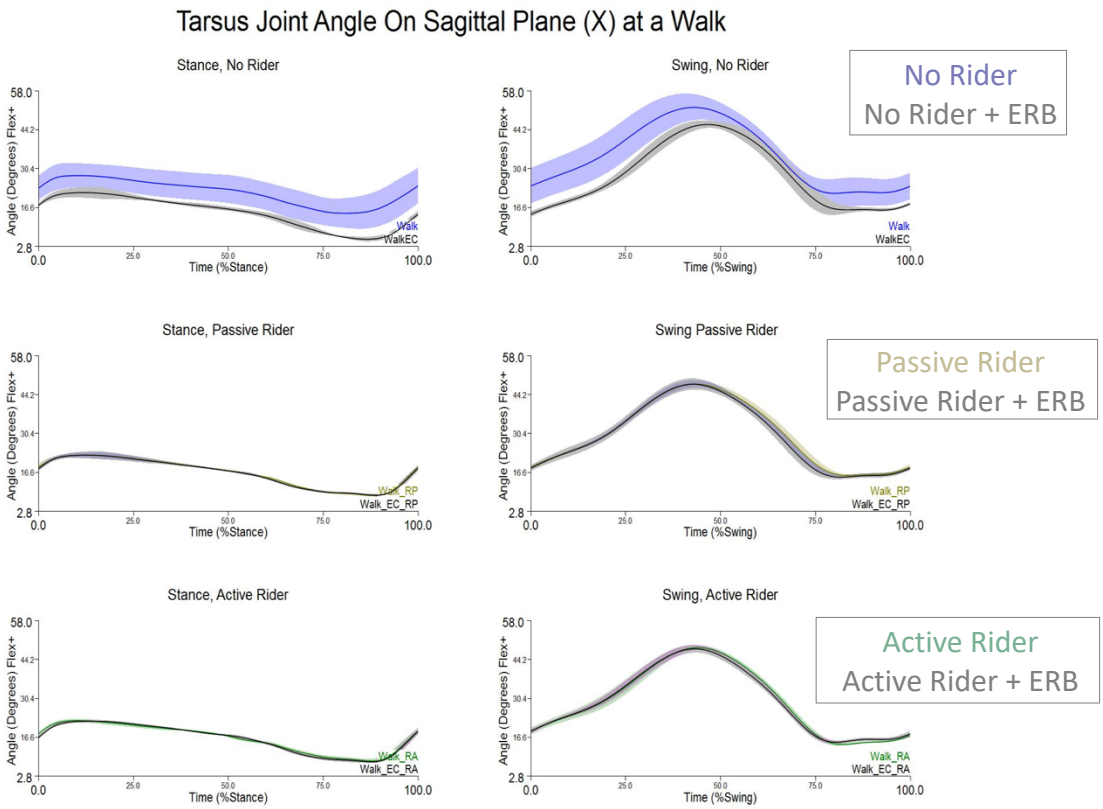


Figure 13. Effect of elastic resistance bands (ERB) on tarsus angle at a walk

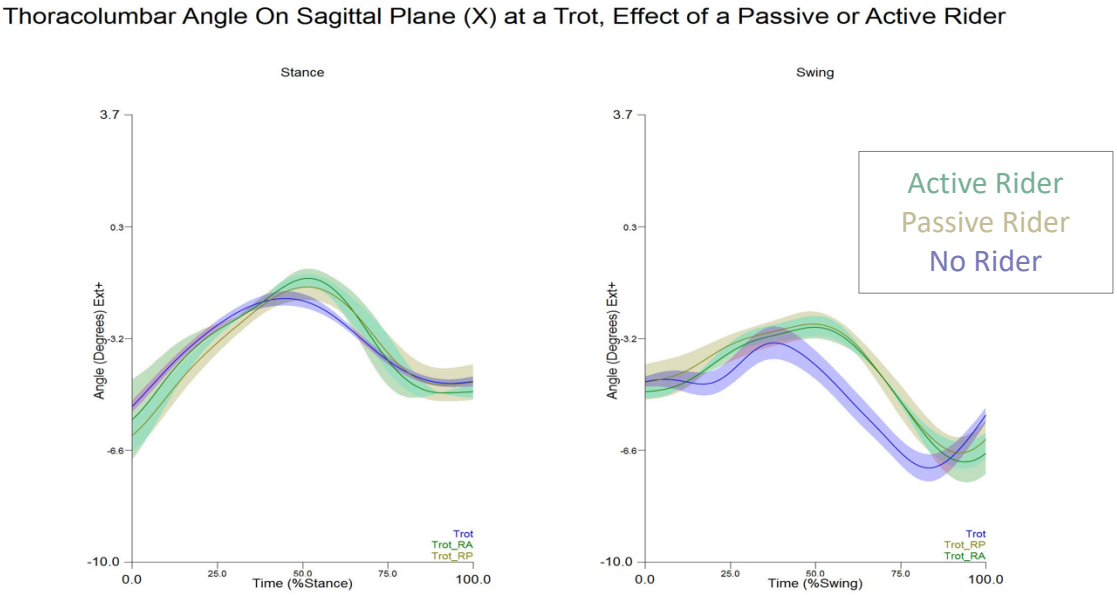


Figure 14. Effect of a “passive” or “active” rider on thoracolumbar angle at a trot.

Table 10: Thoracolumbar (TL) angle at a trot, with different rider conditions (no rider, “passive” rider, and “active rider”), and with and without elastic resistance bands (ERB).

Trot	TL Angle Max		TL Angle Min		TL ROM		TL Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	-1.926	0.221	-7.127	0.415	5.201	0.335	-3.329	0.109
No Rider +ERB	-3.022	0.212	-7.397	0.216	4.375	0.422	-4.724	0.124
Passive Rider	-1.569	0.401	-6.677	0.500	5.108	0.558	-3.408	0.291
Passive Rider +ERB	-3.961	0.642	-9.144	0.925	5.183	1.444	-6.108	0.397
Active Rider	-1.624	0.457	-6.721	0.469	5.097	0.897	-3.552	0.360
Active Rider +ERB	-3.057	0.533	-8.210	0.338	5.153	0.354	-5.192	0.318

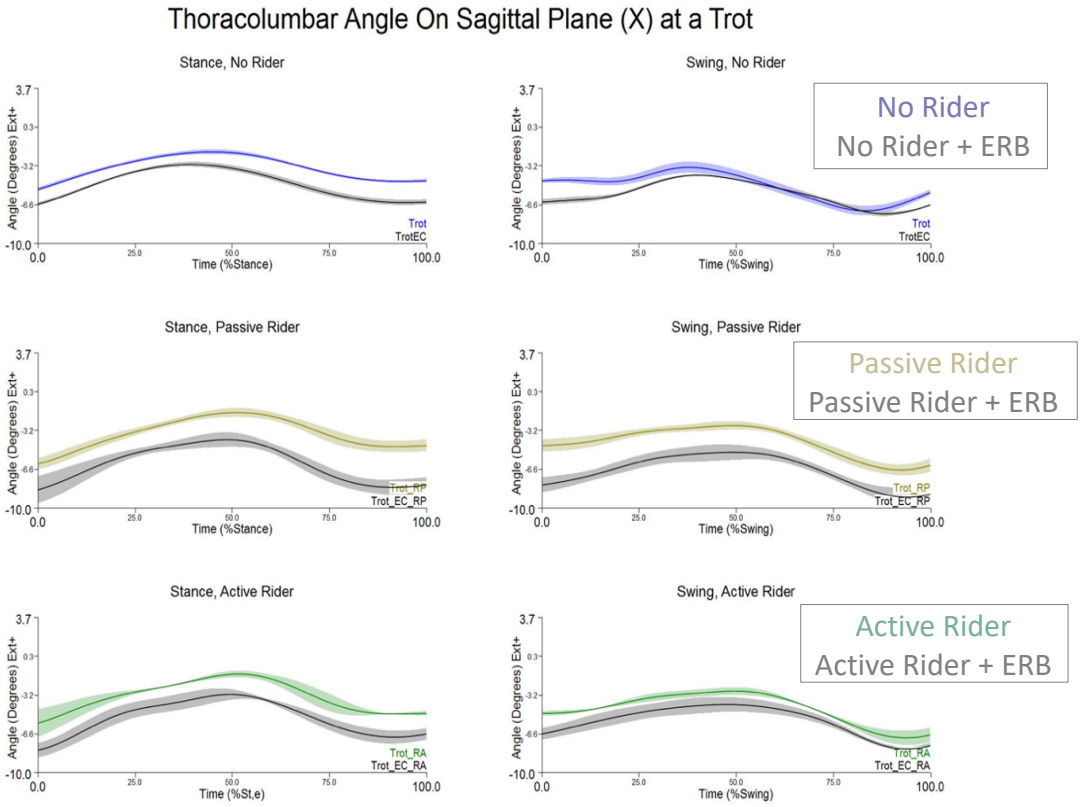


Figure 15. Effect of elastic resistance bands (ERB) on thoracolumbar angle at a trot

relative increase in thoracolumbar joint flexion with the presence of elastic resistance bands increased by up to 2.4 degrees in the “passive” rider condition.

Lumbosacral Angle, At A Trot (Figures 16 & 17, Table 11)

When evaluating the effect of a rider on the lumbosacral joint angle at a trot (Figure 16), a trend is observed with a distinct increase in overall joint flexion observed with both rider conditions compared to with no rider. This increase in overall joint flexion is characterized by no overlap in the standard deviation of the mean joint angles, and a relative increase in maximal flexion by approximately 0.6 degrees in the “passive” and “active” rider conditions compared to no rider. The overall range of motion is slightly reduced in both “active” and “passive” rider conditions as compared to without the rider, however, the standard deviation is also broader in the late stance phase for both rider conditions (Table 11). When evaluating the effect of the elastic resistance bands on the lumbosacral joint angle at a trot (Figure 17), there is no discrete separation of the joint angle curves in the “passive” rider condition with and without elastic resistance bands, however, slight increased joint flexion is seen in both the no rider and active rider conditions with the presence of the elastic resistance bands in the first 75% of stance phase.

Tarsus Angle, At A Trot (Figures 18 & 19, Table 12)

When evaluating the effect of a rider on the tarsus joint angle at the trot (Figure 18), a trend is observed with slight increased overall joint extension observed with both rider conditions compared to no rider. When evaluating the effect of the elastic resistance bands on tarsus angles at the trot (Figure 19), there was no appreciable difference seen with either the “active” or “passive” rider. However, in the condition without the rider, a relative increase in the tarsus extension was seen with the presence of the elastic resistance bands. While increased overall variability was seen in the joint angles without the presence of the rider, the broad shaded standard deviation line was still distinct from the tarsus angle curve with the presence of the elastic resistance band.

Discussion

When evaluating changes in the thoracolumbar and lumbosacral joint angles amongst the various conditions studied at the walk, it is important to note that the scale of these differences in joint angles ranges from approximately 1.5 to 2.3 degree differences in the thoracolumbar joint, and by approximately 0.7 to 0.8 degree differences in the lumbosacral joint. With such small overall changes in joint angles induced by rider or elastic resistance band conditions, this may have minimal clinical significance. However, given that these joints are located on the axial skeleton, a restricted amount of overall joint flexion and extension of the spine is expected. Additionally, when looking at the overall range of motion

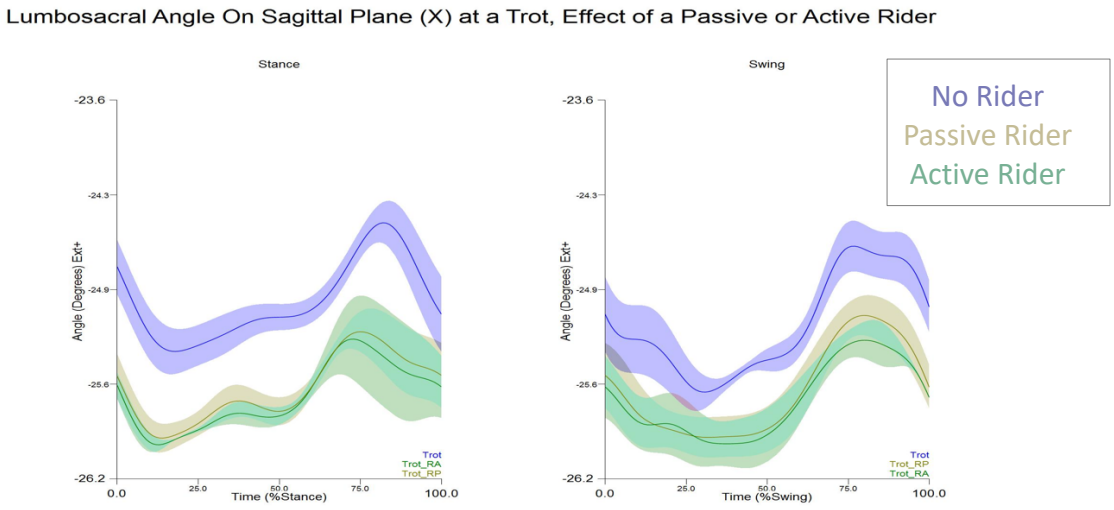


Figure 16. Effect of a "passive" or "active" rider on lumbosacral angle at a trot.

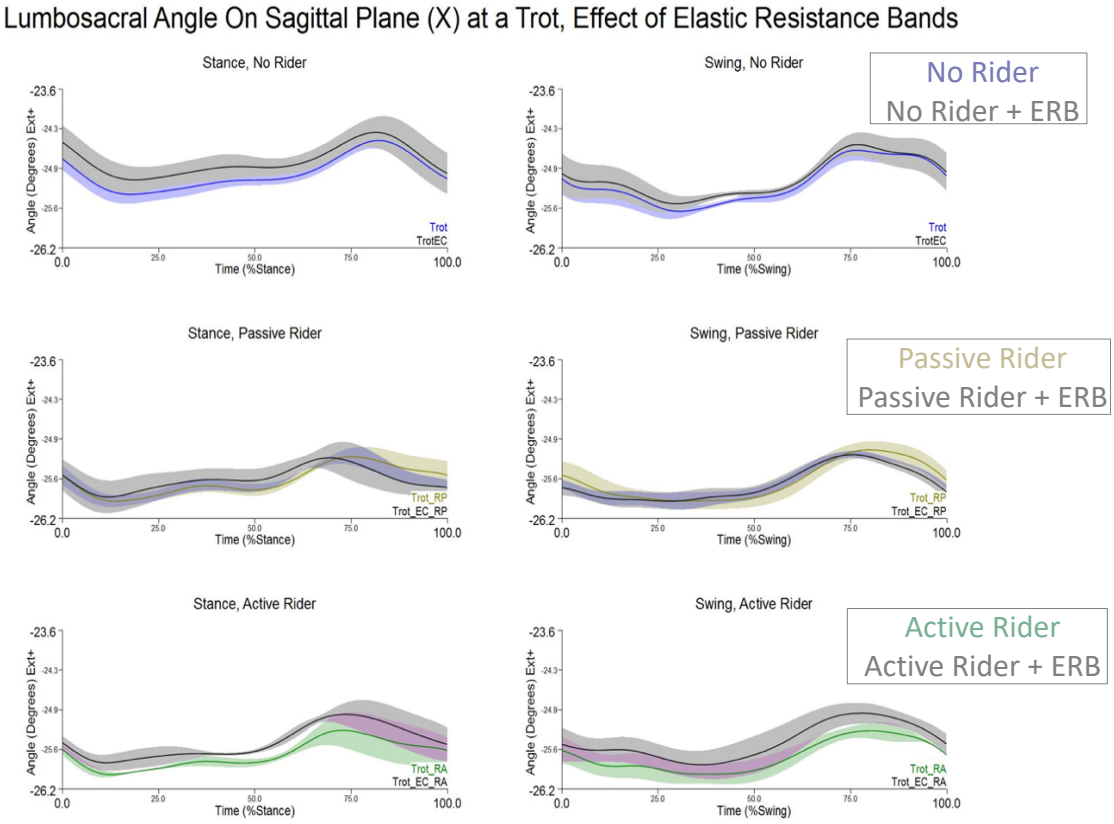


Figure 17. Effect of elastic resistance bands (ERB) on lumbosacral angle at a trot

Table 11: Lumbosacral (LS) angle at a trot, with different rider conditions (no rider, “passive” rider, and “active rider”), and with and without elastic resistance bands (ERB).

Trot	LS Angle Max		LS Angle Min		LS ROM		LS Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	-24.434	0.140	-25.329	0.156	0.895	0.048	-24.981	0.138
No Rider +ERB	-24.296	0.200	-25.190	0.216	0.893	0.110	-24.817	0.186
Passive Rider	-25.171	0.130	-25.929	0.097	0.758	0.074	-25.587	0.109
Passive Rider +ERB	-25.149	0.191	-25.884	0.243	0.735	0.120	-25.561	0.175
Active Rider	-25.099	0.253	-25.901	0.106	0.802	0.274	-25.553	0.121
Active Rider +ERB	-24.937	0.204	-25.788	0.132	0.850	0.265	-25.440	0.086

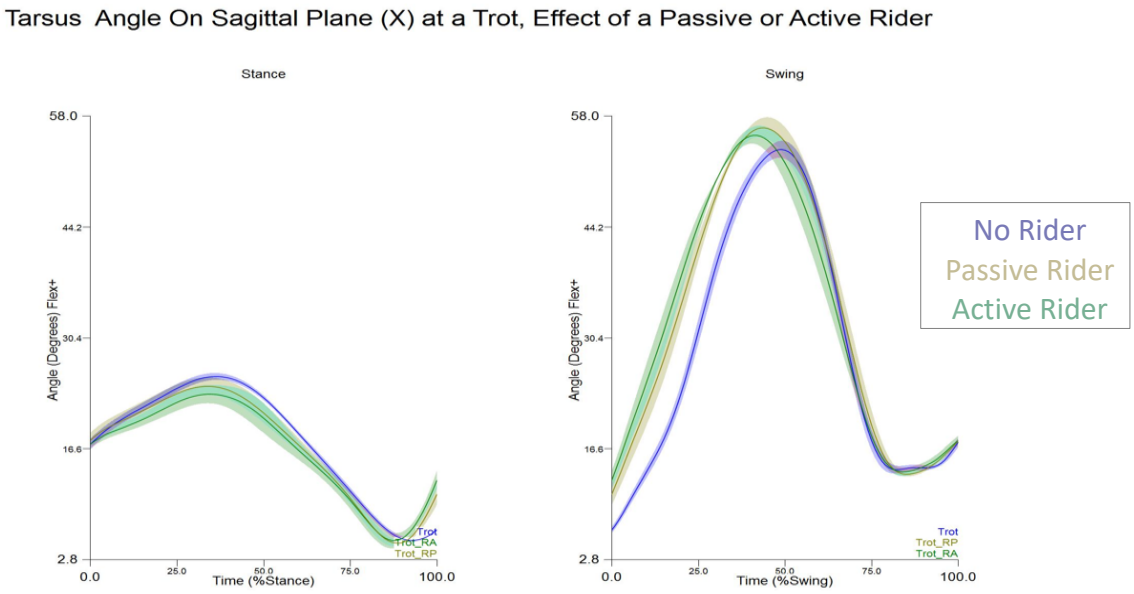


Figure 18. Effect of a “passive” or “active” rider on tarsus angle at a trot.

Table 12: Tarsus angle at a trot, with different rider conditions (no rider, “passive” rider, and “active rider”), and with and without elastic resistance bands (ERB).

Trot	Tarsus Angle Max		Tarsus Angle Min		Tarsus ROM		Tarsus Metric Mean	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
No Rider	53.880	1.041	5.158	0.303	48.722	1.116	17.415	0.355
No Rider +ERB	54.013	0.910	4.700	0.195	49.312	1.077	17.196	0.577
Passive Rider	56.613	1.166	4.753	0.327	51.861	1.352	16.767	0.764
Passive Rider +ERB	55.358	1.874	3.443	0.437	51.915	1.520	15.009	0.949
Active Rider	55.552	0.710	4.865	0.664	50.687	1.201	16.465	0.680
Active Rider +ERB	55.029	1.779	4.550	0.163	50.478	1.756	15.810	0.313

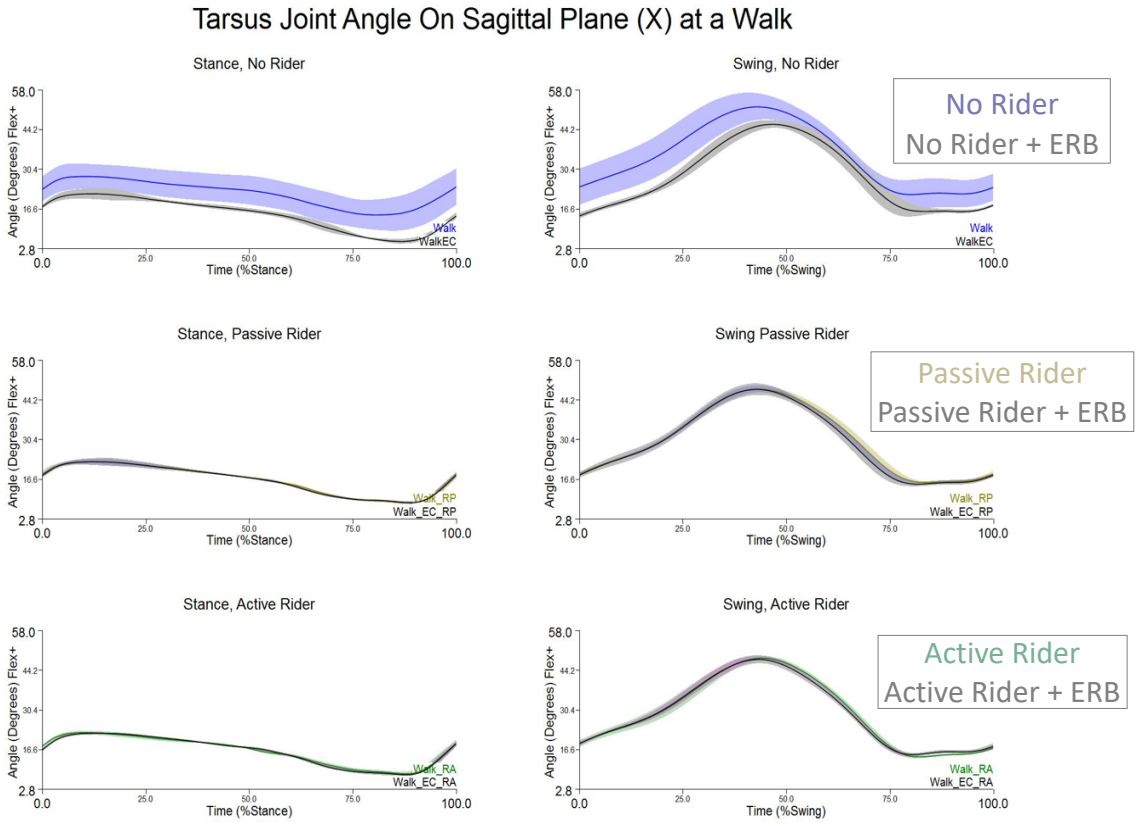


Figure 19. Effect of elastic resistance bands (ERB) on tarsus angle at a trot. of these joints, the largest range of motion seen amongst the various conditions is 8.0 degrees in the thoracolumbar joint and 1.0 degree in the lumbosacral joint.

When considering that the maximal range of motion detected in the lumbosacral joint is 1.0 degree, changes in the joint angle by 0.7 degrees represents 70% of the maximal range of motion detected in that joint, which may represent a clinically relevant change for that particular joint. Similarly, a 1.5 – 2.3 degree difference in the thoracolumbar joint may hold some clinical significance when the maximal range of motion of this joint was detected around 8.0 degrees in the examined conditions.

Since this was a pilot study, one of the aims was to develop a maker set for future joint or segment investigations, and to increase familiarity with the motion capture equipment and data processing. To be applied to larger scale studies, the accuracy of marker placement will become increasingly important for consistency of data collection. This will require consistent identification of anatomical landmarks across various horses, and precision in placement of the reflective markers.

Another factor that is critical in the data analysis stage is the accuracy in defining gait events (heel-strike and toe-off). There have been several previous reports in the literature describing methods of defining heel-strike and toe-off events in quadrupeds in the absence of a force platform [43-45]. One such technique involves utilizing peaks in vertical acceleration to determine heel-strike, and the minimum in vertical velocity to identify toe-off events. While this has been demonstrated in horses, these variables have not been found to be a reliable for identifying gait events accurately in human studies. Therefore, for future application in larger studies, a validation trial to confirm the gait event marking strategy with force data is recommended. Or alternatively, using a larger number of horses to demonstrate consistency in identification of gait events using any particular technique for identifying gait events.

While in this limited pilot study the use of the peak in horizontal velocity of the center of gravity of the left hind foot-pastern segment for identifying heel strike and the minimum in horizontal acceleration of the center of gravity of the LH foot-pastern segment to find the toe off event were found consistent in this one sample subject, repeating this technique on more horses is necessary to demonstrate the accuracy and increase the validity of this technique for identifying gait events. For the purposes of this small study, it was elected to track the center of gravity of the entire left hind foot-pastern segment rather than of any one particular marker, as with multiple markers identifying the segment, this was found to be a more stable marking system than when utilizing individual markers.

For this particular study, the joint angle graphs were produced to demonstrate stance and swing phases, given that it was unknown if the presence of the rider or elastic resistance bands would have had greater effects in either phase of the gait cycle. Typically, the stance phase is when greater forces are applied to the stance limb, and therefore when greater increases in joint angles are observed [33-34]. However, given that quadrupeds have more complex gait patterns with multiple stance limbs depending on the gait under investigation, it

may be worthwhile to continue with joint angle investigation throughout the entire gait cycle until it is better understood.

While several trends were identified in this pilot study with the presence of the rider and elastic resistance bands, it is important to acknowledge that these are not necessarily statistically significant changes, but merely a trend identified with one sample subject. To investigate for statistical significance of these trends, a greater number of sample subjects is needed followed by statistical testing. It is also important to acknowledge that there are certain inherent limitations involved with the use of motion capture. Some of these limitations include the potential for measurement error, as well as error with gap filling. It is important to try to minimize these errors with good technique, an appropriate marker set, good lighting, an increased number of cameras, and appropriately placed and sized reflective markers.

In this particular study, joint angles were limited to the sagittal plane. With the use of 3D motion capture, there is an opportunity for investigation of joint or segment angle changes on different planes, or to investigate angular or rotational movement. Which angles are investigated depends on the scientific questions being posed and on the type of motion being examined.

Additionally, another limitation is that data was only collected on a hard surface, as some horses alter their kinematics on a soft surface where they are more comfortable pushing off on a deformable surface with more traction and less impact. It is possible that some of our kinematic findings may have been different on a soft surface versus asphalt. Another factor when considering hindlimb joint angles is changes in stride length, as a reduction in stride length associated with increased spinal flexion may have an effect on reducing limb flexion angles. This area requires further study, as it has not yet been documented what degree of spinal flexion results in certain changes in stride length or resulting changes in hindlimb joint angles.

The study of kinematics holds vast potential to examine changes in motion of horses with the application of different therapeutic exercise strategies or training tools. Additionally, it could be utilized as a means of demonstrating any effects of alterations in motion after a particular training period. It can also be utilized in conjunction with force data or electromyographic data to provide more detailed information regarding concurrent muscle activation, or information regarding the force, power, or moments involved with different exercise conditions or strategies. Through the utilization of these research techniques, we can aim to improve the evidence base for different therapeutic exercise techniques being used in horses.

Conclusion

Overall, the use of an abdominal and hindquarter elastic resistance band applied at 25% stretch has resulted in a significant increase sEMG activity in the

left rectus abdominus muscles at a trot. Additionally, there is a trend towards a reduction of sEMG activity of the longissimus dorsi muscles at a walk and trot, which was not found to be statistically significant. Additionally, pilot kinematics data in horses instrumented with elastic resistance bands have identified an increase of flexion of the thoracolumbar region at a walk and trot with and without a rider, in addition to increased lumbosacral flexion with the presence of an active rider. To have a “rounded” or “collected” posture, the thoracic spine must lift relative to the pelvis. While the cranial and mid-thoracic spine was unable to be visualized along its dorsal aspect due to the application of the saddle and pad for attachment of the elastic resistance bands, this increase in thoracolumbar joint angle does suggest rounding and flexing of the spine at this level. The presence of elastic resistance bands has also demonstrated increased extension of the tarsus without a rider in one horse at a walk and trot, which is suggestive of increased hindquarter activation in response to the increased proprioceptive stimulus.

The use of ground poles and their influence on sEMG activity have also been investigated. The use of ground poles has resulted in a significant increase in sEMG activity in the rectus abdominus muscles bilaterally at a walk and trot, as well as a significant increase in the sEMG activity of longissimus dorsi muscles bilaterally at the walk.

Pilot data also investigated different rider effects, looking at a “passive” rider versus an “active” rider that is utilizing different seat, leg, and rein aids to influence horse posture and hindquarter engagement. Based on preliminary research trends, increased flexion of the thoracolumbar joint was seen with “active” rider at a walk, and increased flexion of the lumbosacral region was seen in the “passive” and “active” rider at the walk and trot. This effect was greater in the “active” rider as compared to the “passive” rider. There was additionally increased extension of the tarsus seen at the walk and trot with the presence of a rider. Increased extension of the thoracolumbar joint was also identified with the presence of the passive rider as compared to no rider, which was likely due to the weight of the rider on the thoracic spine. However, this effect was reversed with the presence of the active rider, which resulted in increased flexion of the thoracolumbar joint as compared to no rider.

It is also possible that further changes induced by the elastic resistance bands may be a neuromuscular response that may take time to develop. A limitation of this study is that the horses did not have a period of time in training with the elastic resistance bands before data collection, so any kinematic changes or sEMG changes resulting from muscular adaptations over time with training with resistance bands could not be investigated, but this may be a direction for future research.

Clinical Application

Overall, there remains much work to be done to better understand the changes in equine kinematics or electromyographic activity with the application of elastic resistance bands, ground poles, or with the presence of a rider. However significant findings and trends were identified with this research, including (1) increased rectus abdominus and longissimus dorsi activation with the use of ground poles, (2) increased thoracolumbar and lumbosacral flexion and hock extension suggested with the use of elastic resistance bands, and (3) increased thoracolumbar and lumbosacral flexions with the presence of an “active” rider using leg, rein, and seat aids. These findings have identified some benefit to the use of the investigated therapeutic strategies for muscle activation or for the kinematic results described.

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Appendix I: Marker Set

MARKER SET
Elastic Resistance Bands, "Passive" & "Active" Rider

Blue= tracking marker, will be kept on the horse after the calibration
Black= anatomic marker, will be removed after the calibration trial

RHE_2 <Segment 1, Head>
ATLAS_LEFT
ATLAS_RIGHT
EYE_LEFT
EYE_RIGHT

NECK_CR <Segment 2, Cranial Cervical>
CERVICAL4_LEFT
CERVICAL4_RIGHT
C3-4 Cluster
- *Cerv1_1*
- *Cerv1_2*
- *Cerv1_3*
- *Cerv1_4*

ATLAS_LEFT
ATLAS_RIGHT

NECK_CD <Segment 3, Caudal Cervical>
CERVICAL7_LEFT
CERVICAL7_RIGHT
C5-C6 Cluster
- *Cerv2_1*
- *Cerv2_2*
- *Cerv2_3*
- *Cerv2_4*

CERVICAL4_LEFT
CERVICAL4_RIGHT

SHOULDER <Segment 4, Shoulder>
CR_CDBR_SCAPULA_LEFT
CD_CDBR_SCAPULA_LEFT
SCAPULA Cluster
- *SCAP1*
- *SCAP2*
- *SCAP3*
- *SCAP4*

SUPRAGLENOID_TUBERCLE_LEFT

LAR_2 <Segment 5, Humerus>
SHOULDER_LEFT (Greater Tubercle Humerus)
HUMERUS Cluster
- *HUM1*
- *HUM2*
- *HUM3*
- *HUM4*

DISTAL_HUMERUS_LEFT (Supracondylar Crest)

LFA_2 <Segment 6, Radius>
ELBOW_LEFT (Olecranon Process)
RADIUS Cluster
- *RAD1*
- *RAD2*
- *RAD3*
- *RAD4*

CARPUS_LAT_LEFT
CARPUS_MED_LEFT

LEFT_HAND <Segment 7, LF Metacarpus>
CARPUS_LAT_LEFT
CARPUS_MED_LEFT
METACARPUS Cluster
- *MC1*
- *MC2*
- *MC3*
- *MC4*

FETLOCK_FORE_LAT_LEFT
FETLOCK_FORE_MED_LEFT

LEFT_FORE_HOOF <Segment 8, LF Pastern-Foot>
FETLOCK_FORE_LAT_LEFT
FETLOCK_FORE_MED_LEFT
PASTERN Cluster
- *FPAS1*
- *FPAS2*
- *FPAS3*
- *FPAS4*

CORONARY_FORE_LAT_LEFT
CORONARY_FORE_MED_LEFT

RTA_2 <Segment 9, Thoracic>
LUMBAR 1
SADDLE PAD Cluster Left
SADL1
SADL2
SADL3
SADL4
SADDLE PAD Cluster Right
SADR1
SADR2
SADR3
SADR4
CERVICAL7_LEFT
CERVICAL7_RIGHT

LUMBAR_SPINE <Segment 10, Lumbar>
LUMBAR1
LUMBAR Cluster
LUM1
LUM2
LUM3
LUM4
SACRUM_JOINT
Anterior orientation - Virtual Anatomical Marker: mid tuber coxae, (50% between left and right tuber coxae)

SACRUM <Segment 11, Sacral>
SACRUM_JOINT
Anterior orientation - Virtual Anatomical Marker: mid tuber coxae
SACRUM Cluster
SAC1
SAC2
SAC3
SAC4
SACRAL5

RPV_2 <Segment 12, Pelvis>
TUBER_COXAE_LEFT
TUBER_COXAE_RIGHT
PELVIS Cluster
PEL1
PEL2
PEL3
PEL4
CROUP_LEFT (Mid-Greater Trochanter, over coxofemoral joint center)
CROUP_RIGHT

LTH_2 <Segment 13, Femur>
CROUP_LEFT (Mid-Greater Trochanter, over coxofemoral joint center)
(CROUP_RIGHT)
FEMUR Cluster
FEM1
FEM2
FEM3
FEM4
KNEE_LAT_LEFT(Lateral Stifle Joint; above tibial plateau)
KNEE_MED_LEFT

LSK_2 <Segment 14, Tibia>
KNEE_LAT _LEFT (Lateral Stifle Joint; above tibial plateau)
KNEE _MED _LEFT
Tibial Cluster
TIB1
TIB2
TIB3
TIB4
TARSUS_LAT_LEFT
TARSUS_MED_LEFT

LFT_2 <Segment 15, Metatarsus>
TARSUS_LAT_LEFT
TARSUS_MED_LEFT
METATARSUS Cluster
MTT1
MTT2
MTT3
MTT4
FETLOCK_HIND_LAT_LEFT
FETLOCK_HIND_MED_LEFT

LEFT_HIND_HOOF <Segment 16, LH Pastern-Foot>
FETLOCK_HIND_LAT_LEFT
FETLOCK_HIND_MED_LEFT
PASTERN Cluster
HPAS1
HPAS2
HPAS3
HPAS4
CORONARY_HIND_LAT_LEFT
CORONARY_HIND_MED_LEFT

VITA

Karen Shaw graduated from veterinary school at the University of Glasgow, Scotland in 2011. Following which she completed a one-year rotating equine internship at North Carolina State University, followed by a one-year equine surgery and lameness internship at the University of California, Davis. Karen then began an Equine Sports Medicine and Rehabilitation Residency at The University of Tennessee, Veterinary Teaching Hospital, where she did a dual Masters' Program in Comparative And Experimental Medicine (CEM).