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Specificity and variability of trunk kinematics on a mechanical horse

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ABSTRACT

As perturbation training is gaining popularity, it is important to better understand postural control during complex three-dimensional stimuli. One clinically relevant and commonly used three-dimensional stimulus is found in hippotherapy and simulated hippotherapy on a mechanical horse. We tested nine healthy participants on a horse simulator, measured head and trunk kinematics, and characterized data in time (root-mean-square and variability) and frequency (amplitude spectra, gains, and phases) domains. We addressed three fundamental questions: 1) What is the specificity of postural responses to the simulator? 2) Which plane of motion is associated with the most and least variability (repeatable movements across repeated stimuli and across participants)? 3) To what extent are postural responses influenced by different degrees of stability (addition of pelvis straps and trunk support)? We found head and trunk responses were highly specific to the three-dimensional simulator perturbation direction and frequency. Frontal plane responses had the least variability across repetitions and participants whereas transverse motion was most variable. Head motion was more variable than the trunk at low frequencies and exhibited a marked decrease in tilt in the sagittal plane. Finally, the inclusion of pelvis straps had minimal effect on kinematics at low frequencies but altered higher frequencies; whereas added trunk support reduced head and trunk responses to perturbations and altered timing characteristics in all three planes. In conclusion, the present study suggests that frontal plane motion was under a high level of control, and results support the idea that specific head and trunk postural responses can be elicited from a complex three-dimensional stimuli, such as those found in hippotherapy. Researchers and clinicians can use results from this study to help interpret variability, implement mechanical adjustments to stability, and assess responses in pathological populations.

1. Introduction

Trunk control is foundational to posture and most voluntary activities (Edwards, Austin, & Bird, 2017; Karthikbabu et al., 2012; Rachwani, Santamaria, Saavedra, & Woollacott, 2015). Impairments in motor control of the trunk are common in a variety of orthopedic (Mokhtarinia, Sanjari, Chehrehrazi, Kahrizi, & Parnianpour, 2016) and neurologic conditions, such as stroke (Karthikbabu et al., 2012), Parkinson's Disease (Maetzler et al., 2012), spinal cord injury (Harel et al., 2013; Seelen, Potten, Huson, Spaans, & Reulen, 1997), and cerebral palsy (Saavedra & Woollacott, 2015). Hippotherapy and simulated hippotherapy on a mechanical horse are useful tools to improve trunk postural control in multiple sclerosis (Bronson, 2010), cerebral palsy (Dewar, Love, & Johnston,

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2015), spinal cord injury (Lechner, Kakebeeke, Hegemann, & Baumberger, 2007), and even insulin sensitivity (Kubota et al., 2006). The primary advantages of a mechanical horse include enhanced safety, lower chance of rider fear, and convenient indoor access for increased frequency of riding sessions. One reason riding a mechanical horse may improve postural control is because it requires a high number of balance corrections practiced during a typical riding session (Park & You, 2018). Also, if a mechanical horse can accurately simulate live horse motion, then another benefit may be improved gait patterns due to the remarkable similarity in three-dimensional pelvis motion between human gait and riding a horse (Garner & Rigby, 2015).

Despite growing evidence that horse riding motion is beneficial, there remain many fundamental questions about how the trunk posture system responds to such complex, three-dimensional motion stimuli. Understanding these responses should be valuable for the design, improvement, and assessment of therapeutic interventions. Further, as wireless technology is making it less intrusive and more accessible to monitor rider kinematics (Chen, Lach, Lo, & Yang, 2016; De Vito, Postolache, & Rapuano, 2014), the need to understand and properly interpret postural responses is ever more pressing. The present study incorporates a series of experimental trials on a mechanical horse-riding simulator to address three foundational questions about the nature of human kinematic postural responses and the influence of biomechanical modifications that change the inherent stability of the rider.

First, what is the specificity of postural responses to horse riding motions? We determine if horse motion in a particular plane (sagittal, frontal, and transverse) and at a particular stimulus frequency translates to linear postural responses (i.e., tilt or rotation response in the same plane and same frequency as horse motion). Previous research in standing conditions generally show high specificity in postural responses to stimuli in a single plane of motion (Goodworth & Peterka, 2009; Kiemel, Elahi, & Jeka, 2008; Peterka, 2002; Mergner, Schweigart, Maurer, & Blumle, 2005; van der Kooij & Peterka, 2011) and in response to two different stimuli in different planes of motion (Hwang, Agada, Kiemel, & Jeka, 2014; Cenciarini & Peterka, 2006; Mergner et al., 2005). However, no study has detailed the linearity across planes of motion and frequency in response to a three-dimensional stimulus, which represent everyday situations (e.g., responding to a bump that simultaneously rotates and pushes you, walking over a bumpy surface). In contrast to the notion of high specificity, there is evidence for non-linearity throughout all stages of a postural response: variable muscle stiffness (Loram, Maganaris, & Lakie, 2007), noise in sensory interpretations (Faisal, Selen, & Wolpert, 2009), threshold-based control (Gawthrop, Loram, Lakie, & Gollee, 2011), and the neural signals that activate muscles (Preuss & Fung, 2008). In addition, trunk muscles cross many spinal joints, are oriented across planes of motion (Moore & Agur, 2010), and activate during both frontal and sagittal plane stimuli (Preuss & Fung, 2008). In response to three-dimensional surface stimuli, a single muscle may be used to simultaneously respond across all three planes of motion. Therefore, it is possible that trunk postural responses to complex three-dimensional horse motion be distributed across frequencies unrelated to the horse motion, or that responses include sizable interactions across planes of motion. For example, horse motion in the transverse plane may produce postural responses in all three planes of motion. However, if postural responses are highly specific to horse motion, then researchers and clinicians can select complex motion profiles to trigger postural responses at specific frequencies and directions. Moreover, comparing the degree of linearity in postural control in patients with impaired control to that in healthy adults can be a valuable tool for assessment of pathology and tracking progress in hippotherapy or on a mechanical horse.

Second, which plane of motion is associated with the most variability? We characterize variability across repeated stimulus cycles and across participants and compare this variability across frontal, sagittal, or transverse planes. While variability is important to quantify in motor control, there is no agreed upon interpretation (Komar, Seifert, & Thouvarecq, 2015). Some argue variability in performance is minimized as an individual increases skill level (Schmidt, 2003), while others posit that variability is inherent to complex systems and should be investigated as a clue to transitions in motor learning and interactions between neural, biomechanical, and environmental factors (Newell & Corcos, 1993; Magill, 2014), and still others suggest variability within individual body segments is a positive sign as long as variability in the overall movement goal is sufficiently low (Latash, Scholz, & Schöner, 2002). With this third view in mind, we consider that a body segment whose motion or biomechanical interaction plays little role in the overall movement goal may exhibit greater variability than segments more directly linked to that goal. In the present study, the differences in trunk musculature, anatomy, and base of support across the three planes suggest that a rider may allow greater variability in one direction while suppressing variability in other directions. Thus, measuring variability may help understand which body segments and movements are prioritized.

Third, how do postural responses change with various levels of enhanced stability? We reexamine the relation between postural responses and riding motion stimulus after altering stability by providing pelvis straps and trunk support to riders during select trials. These modifications are modeled after the Segmental Assessment of Trunk Control, a National Institutes of Health recommended posture assessment tool (NIH 2018) and are increasingly used in populations with severely impaired trunk control who cannot sit independently (Saavedra & Woollacott, 2015; Butler, Saavedra, Sofranac, Jarvis, & Woollacott, 2010). For example, in children with severe CP, pelvis and trunk support is used to quantify partial trunk control (Butler et al., 2010) and have been shown to improve posture and upper extremity function (Rachwani et al., 2015). Interestingly, for those with severe CP, hippotherapy and mechanical horse riding provide some of the best evidence for improvement (Dewar et al., 2015). In general, postural control is dependent on the body mechanics (Bingham, Choi, & Ting, 2011; Cordo & Nashner, 1982), and even relatively small changes can affect postural control, such as different saddle shapes as may be experienced during hippotherapy (Ribeirom, 2018). Therefore, to understand postural responses when a patient requires pelvis or trunk support, it is necessary to first understand how these modifications alter postural control in healthy riders.

The present study is the first to address fundamental questions that relate three-dimensional stimulator motion during mechanical horse riding with trunk postural control. Many of the study conclusions should also be important for clinicians and researchers working more generally with perturbations.

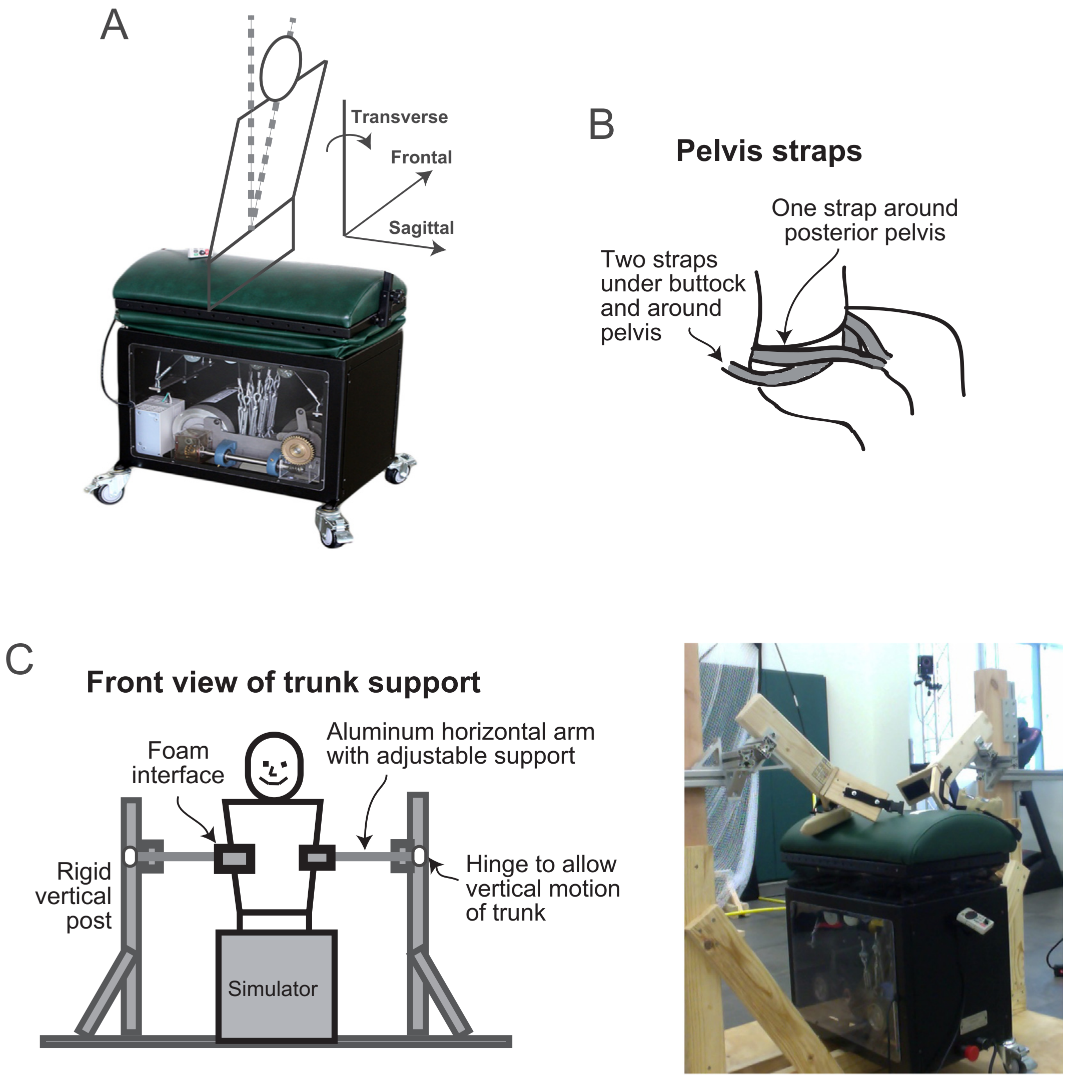

Fig. 1. A) Image of mechanical horse (ie, simulator) and reference frame. B) Image of pelvis straps and C) trunk support schematic and photograph. The pelvis straps and trunk support were used to modify stability in certain trials.

2. Methods

Motion capture technology (Vantage, Vicon Motion Systems, LTD, Oxford, UK) was used to record angular kinematics during human riding trials on a mechanical horse riding simulator. Trials included variations with and without different forms of external support. Analyses of angular kinematics were performed to assess and compare trunk responses of the participant riders to the input stimuli of the mechanical horse.

2.1. Participants

Nine healthy participants, six male and three female, were included in this study. The participant pool had a mean age of 24.5 years ± 9.3, height of 170.2 cm ± 15.6, and mass of 68.0 kg ± 17.4. Two participants had previous experience with the mechanical horse (< 20 min) and one participant had experience on a live horse. All participants were tested according to a protocol approved by the Baylor University Internal Review Board (IRB) and gave their informed consent.

2.2. Mechanical horse-riding simulator

Each participant completed a series of trials riding on a mechanical horse simulator that elicits three-dimensional motion similar to that experienced when riding a horse (Fig. 1A). The mechanical horse (hereafter referred to as the simulator) is a stationary

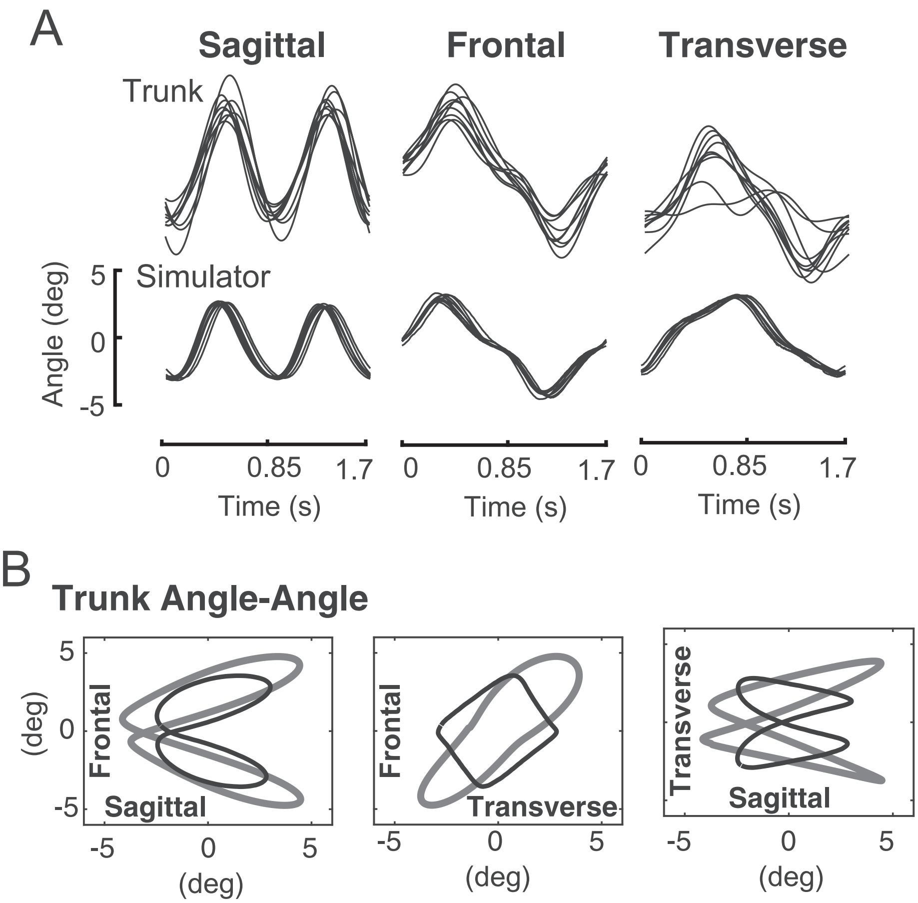

Fig. 2. A) Sample data of trunk motion from all participants from one control trial, B) Angle-angle diagram showing the relative timing of simulator (black) and human trunk response (grey) between planes of motion.

apparatus with a seat structure supported by a set of eight cables and driven by a set of eight cams (Benoit, 2011). The cams are shaped to produce a target motion pattern in the seat that was derived from three-dimensional motion-capture recordings of a live horse at a normal walking gait. The simulator is driven by a variable-speed motor set for this study at a single, constant rate of 1.71 s per cycle for all trials (0.585 strides per second), which is typical of a horse walk. The seat structure is stabilized by springs that allow for small variations in seat amplitude to occur from cycle to cycle, depending on participant size and responses. Fig. 2A shows the three-dimensional angular kinematics of one simulator motion cycle averaged over one riding trial from all nine participants. Though the focus of this study is postural responses referenced to the angular kinematics of the simulator (i.e., in the sagittal plane, in the frontal plane, and in the transverse plane), the simulator seat also generates small translational motions which may influence the trunk responses, as described in more detail in the Discussion.

2.3. Protocol

Prior to data collection, all participants in this study were familiarized with the simulator (Fig. 1A) by riding for five minutes. After a rest period of 5 min, each participant completed a sequence of four one-minute riding trials: one control trial with no external support, followed by one pelvis strap trial, followed by one trial with both pelvis strap and trunk support (pelvis + trunk support), followed by a second control trial at the end. For the control trials, the participants sat comfortably, arms by their side, and facing forward with eyes open. All participants had their feet off the ground and feet to floor distances ranged from about 5 cm to 20 cm, which was unlikely to differentially introduce any fear of falling in some subjects more than others. Participants were instructed to sit upright and respond naturally.

Trials with pelvis straps were identical, but included comfortable straps around the pelvis to stabilize the pelvis and orient it more vertically (Butler et al., 2010), Fig. 1B. The pelvis straps helped align the pelvis to the seat in all planes of motion. But the straps were especially designed to impact pelvis orientation in the sagittal plane by reducing posterior pelvis tilt (which is present in some populations with impaired motor control). Trials with the trunk support included a padded stabilizing support located around the lower ribs (Fig. 1C). The trunk support reduced horizontal translations at the point of contact (Goodworth, Wu, Felmlee, Dunklebarger, & Saavedra, 2017) but allowed vertical motion on either side of the support up and down. Sagittal and transverse plane motion was also reduced at the point of contact (see triangular contact piece in photo in Fig. 1C). Together, the trunk support stabilized spinal segments below the level of support against gravity. Adding pelvis straps and then trunk support progressively added more stability to the upper body; and these modifications are clinically important (Butler et al., 2010; Saavedra & Woollacott, 2015).

Trials for each participant were completed during a single session in the following order: first control trial, then pelvis strap trial, then pelvis + trunk support trial, and then the second control trial. Due to the setup of pelvis straps and trunk support, it was not convenient to randomize the order of trials. However, nearly all dependent variables of trunk kinematics across the three planes of motion proved not significantly different between the 1st and 2nd control trial, suggesting that learning was minimal within the session. A previous study similarly showed that variability in saddle pressures under the rider did not change within a single session of hippotherapy (Janura, Peham, Dvorakova, & Elfmark, 2009).

2.4. Kinematic capture & analyses

A fourteen-camera, infra-red Vicon Vantage motion capture system was used to measure three-dimensional positions of four reflective markers on the simulator (one near each seat corner) and nine markers on the participant (bilateral PSIS, C7, bilateral shoulders, and four equally spaced around a head band). From these markers, kinematic angular positions and velocities of the simulator and participant were calculated using custom Matlab programs (The Mathworks, MA, USA). Postural responses were calculated and included trunk and head angles in the sagittal and frontal planes and trunk and head rotation in the transverse plane (Figs. 1A and 2A). Trunk tilt angles (angular position and velocity with respect to vertical) were computed in each plane as the arc tangent of the horizontal displacement of C7 with respect to the mid-point between PSIS markers divided by the trunk distance between C7 and that mid-point (Goodworth & Peterka, 2009). Trunk rotation angle (angular position and velocity) was defined as the angle of the line connecting shoulder markers in the transverse plane. Similarly, head tilt (angular position and velocity) was defined as the arc tangent of the horizontal displacement of the mid-point of the head with respect to C7 divided by the head segment length (ie, distance between mid-point of head and C7).

Dependent variables included time and frequency domain measures. Time domain measures of trunk and head motion in all three planes included zero-mean root-mean-square (RMS) position and RMS velocity of the average angular response across simulator cycles, the variability across cycles (first calculating the frame by frame standard deviations across cycles and averaging across frames), and coefficient of variation (variability across cycles divided by mean amplitude of response). RMS values quantify overall movement: higher RMS and RMS velocity correspond to higher overall angular positions and velocities. RMS position and RMS velocity were selected because they have an intuitive interpretation, are less sensitive to aberrant movements that impact alternative measures such as peak-to-peak, and are relatively less correlated with each other compared to many alternative measurements (Maurer and Peterka, 2005). Variability quantifies how different responses are across repeated cycles, with higher values indicating more difference. However, since larger overall responses are often associated with larger variability in motor control, the coefficient of variation normalizes variability by the mean response. Frequency domain measures provided more detail into the kinematic responses by decomposing simulator and body segment responses into specific frequencies of motion. Amplitude spectra were calculated as the magnitude of the discrete Fourier transform of simulator and body segment responses. Gains and phases were calculated from the Fourier transforms and indicate the relative magnitude and timing between body segment response and simulator motion at each frequency. For example, a gain of one in the trunk indicates that trunk tilt or rotation was equal in magnitude to the simulator at the particular frequency. Conversely, a gain of zero indicates the trunk stayed upright and was not impacted by the simulator at a particular frequency. A phase of zero indicates the trunk moved with no delay in response to the simulator, whereas a negative phase indicates a lag between response and simulator motion. Gains and phases were calculated at the two frequencies in each plane of motion where power in simulator motion was largest.

2.5. Statistics on experimental data

Each time domain trunk dependent variable (RMS position, RMS velocity, CV, and variability) was analyzed with a separate two way repeated measures ANOVA . Main effects included plane of motion (sagittal, frontal, and transverse) and trial condition (control, pelvis straps, and pelvis + trunk support), and an interaction effect was included between plane of motion and trial condition. Post-hoc pairwise comparisons were made with the Tukey's Honest Significant Difference test. Trunk frequency domain dependent variables (gains and phases) include 95% confidence intervals on the mean values. Each time domain head dependent variable in the control trial was analyzed with a separate one way repeated measures ANOVA with the main effect as plane of motion. Paired t-tests were used to test for differences between head and trunk kinematics and between the first and last control trial. In all statistical analyses, significance was considered at $p < 0.05$.

3. Results

The Results section is organized around the statistical analyses: time domain trunk variables, then frequency domain trunk variables, and then head variables. The first research question (specificity of postural responses) is addressed with the statistical analysis of time domain trunk variables RMS position and RMS velocity across planes of motion, and with the frequency domain analysis. The second research question (variability across body segments and planes of motion) is addressed with statistical analyses of trunk and head dependent variables CV and variability. The third research question (impact of pelvis and trunk support) is addressed with the statistical analysis of trial type across all dependent variables and the differences in frequency domain across

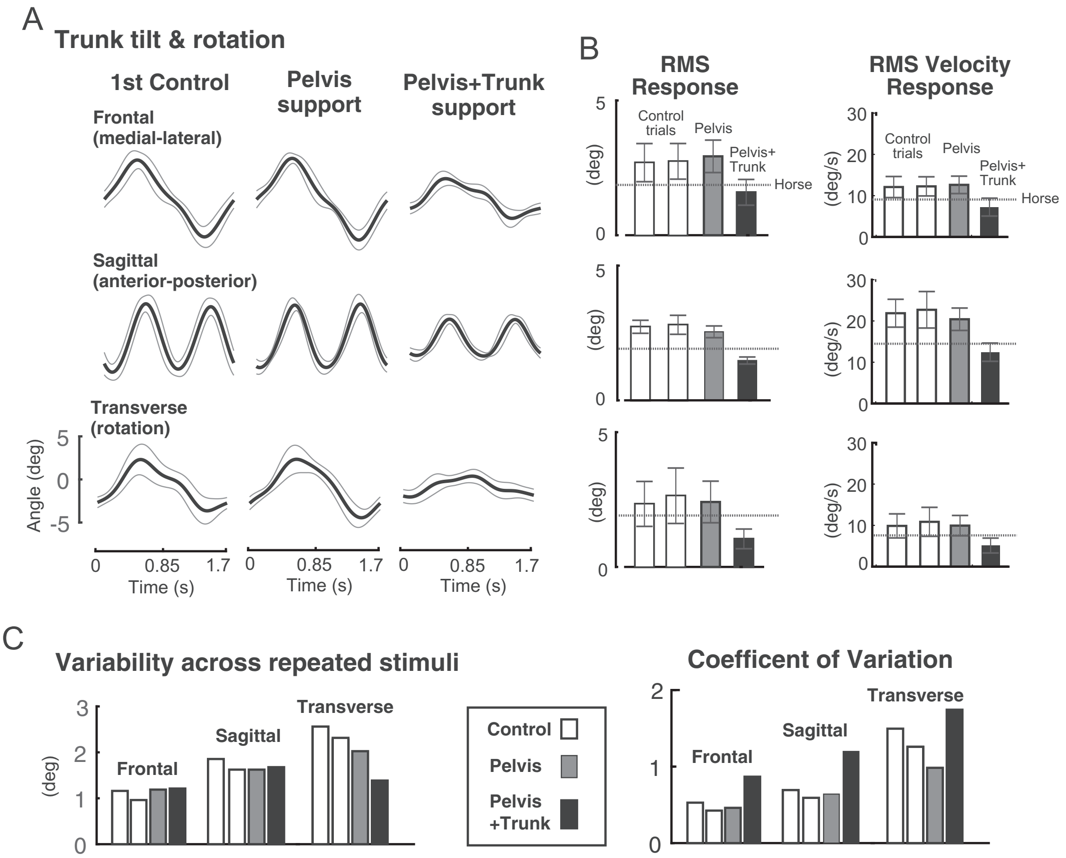

Fig. 3. A) Trunk tilt waveforms averaged across participants with 1 SD across participants for each trial. B) Mean root-mean-square (RMS) position and RMS velocity tilt and rotation with 1SD across participants. C) Average subject variability across cycles and coefficient of variability (CV). See Methods for calculation details.

trials.

3.1. Trunk responses in the time domain

Fig. 2A shows the average cycle of trunk tilt and rotation responses to the simulator motion during a control trial. Each line represents one participant. The uniformity across participants is notable, especially in the frontal and sagittal planes, and was present across all trials. Despite motion across all three planes occurring simultaneously, when separating into individual planes, there is high similarity between simulator and rider response. Fig. 2B shows the timing relation between planes of motion for the simulator and participant response in angle-angle plots. While the simulator motion in Fig. 2A and 2B are displayed across specific planes for clarify, it is important to appreciate that motion occurred simultaneously across all three planes. Participants' responses mimicked the complex simulator pattern with tilt and rotations that often exceeded the stimulus in magnitude.

Across all the trials, average subject RMS position exceeded the simulator in the control and pelvis trials (white and grey bars in Fig. 3B left column), but were less than the simulator with pelvis + trunk support (black bars in Fig. 3B column). Plane of motion had a significant effect on RMS position ($p < 0.0001$) where RMS position was significantly lower in the transverse plane compared to both frontal and sagittal planes. Trial type also had a significant effect on RMS position ($p < 0.0001$), with pelvis + trunk support condition significantly lower than both control and pelvis conditions (Fig. 3A, B). There was not a significant interaction effect between plane of motion and trial type on RMS position.

RMS velocity is shown in Fig. 3B right column. Plane of motion had significant effects on RMS velocity ($p < 0.0001$). All three planes of motion were significantly different from each other, with sagittal greater than frontal and frontal greater than transverse. Although the velocity in the sagittal plane was notably high, the ratio of trunk velocity to simulator velocity (bars divided by horizontal line) remained similar across planes. The average trunk to simulator velocity ratio across participants was 1.35, 1.52, and 1.35 for frontal, sagittal, and transverse planes, respectively, in control trials. The sagittal plane corresponds to the direction of higher frequency of simulator stimulus motion (i.e., one anterior-posterior tilt for each leftward tilt and each rightward tilt), even as RMS position amplitudes in the sagittal and frontal planes are similar. Trial type also had a significant effect on RMS velocity where pelvis + trunk support was significantly lower than both control and pelvis conditions. There was a significant interaction between

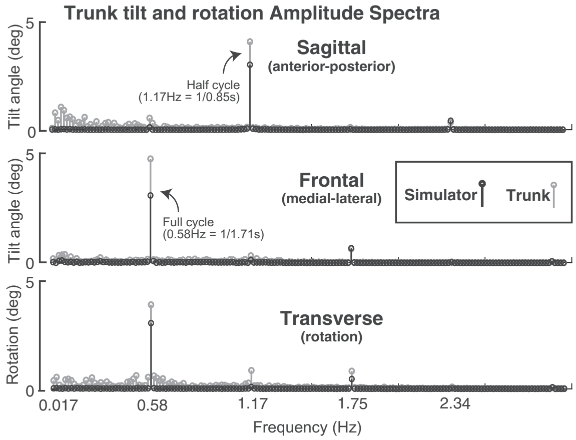

Fig. 4. Sample frequency domain analysis (amplitude spectra) across the three planes of motion for the simulator (black) and human trunk response (grey) for one control trial in one representative participant. In this analysis, the Fourier transform was applied to the entire 60 s cycle to obtain a small fundamental frequency (~1/60 s = ~0.017 Hz) and show motion at low frequencies.

plane of motion and trial type ($p = 0.005$).

Variability results are presented in Fig. 3C left. Plane of motion had a significant effect on variability ($p < 0.0001$) where frontal plane was significantly lower than both sagittal and transverse planes. Trial type also had a significant effect on variability ($p < 0.013$) with trunk + pelvis significantly lower than control condition but not different than pelvis condition. There was a significant interaction between plane of motion and trial type on variability. The influence of trial type had a larger effect in the transverse plane compared to other planes, where the pelvis + trunk condition resulted in a notable reduction in variability.

Results for CV are presented in Fig. 3C right. Plane of motion had a significant effect on CV ($p < 0.0001$) where transverse plane was significantly higher (1.38 averaged across trial) than both sagittal (0.80 average) and frontal (0.58 average) planes. Trial type also had a significant effect on CV ($p < 0.0001$) with trunk + pelvis significantly higher than both control and pelvis conditions. The pelvis + trunk support (black bar) was about 1.6 times the control conditions. There was no significant interaction between plane of motion and trial type.

3.2. *Trunk responses in the frequency domain*

Amplitude spectra quantify trunk responses to simulator tilts and rotations across different frequencies. Fig. 4 shows a sample simulator and trunk response across a 60 sec trial (~0.017 Hz resolution). The amplitude of simulator motion (black dots) was present at distinct frequencies for each plane of motion. In the sagittal plane, simulator tilt was present at ~1.17 Hz and 2.34 Hz. In the frontal and transverse planes, simulator tilt and rotation were present at 0.585 Hz and 1.75 Hz. Trunk motion (grey dots) was present at low frequencies, but was also highly specific to the simulator. In the sagittal plane, notable trunk tilt occurred at 1.17 Hz and 2.34 Hz similar to the simulator, but responses were minimal at frequencies where frontal and transverse plane simulator motion was dominant. In Fig. 4, the slight transverse plane response at 1.17 Hz suggests that in this particular trial there was a minor interaction between planes of motion since the 1.17 Hz corresponds to sagittal plane simulator motion, not transverse.

Amplitude spectra averaged across participants is shown in Fig. 5A. This figure includes the first five harmonic frequencies (integers of 0.585 Hz) where the average simulator cycle had the most power. For example, the first 4 harmonics of the average simulator cycle contained 97.5%, 98.3%, and 97.1% of the total amplitude across frontal, sagittal, and transverse planes, respectively. Across all participants, it is evident that the frequencies where the simulator motion occurred in each plane corresponded to the frequencies where trunk responses occurred. While there was some trunk motion at frequencies where the simulator tilt or rotation in the same plane was small (most notably the transverse response at 1.17 Hz), this trunk motion was quite small compared to the responses elicited by simulator motion within each plane.

To quantify this behavior, we calculated the average ratio of trunk amplitude to simulator amplitude across the two frequencies with the largest simulator motion in each plane. When this ratio was calculated within the same plane, it is termed the "main response" and is equivalent to the average gain between the two frequencies. When this ratio is calculated between planes, it is termed the "interaction response". Main responses in the sagittal, frontal, and transverse plane were similar at 1.26, 1.28, 1.36, respectively, in the control conditions. Trunk tilt in the sagittal plane relative to simulator tilts in the sagittal plane (main response) were approximately 6.5 times larger than interaction responses in the control trial, and 8.7, and 7.3 times larger for the pelvis straps trial and pelvis + trunk support trial, respectively. Similarly, the main response in the frontal plane was about 9.8 times larger compared to the interaction response in the sagittal plane control trial (10.1, 5.7 times larger for pelvis straps and pelvis + trunk support, respectively). The main response in the transverse plane was about 6.2 times larger compared to the interaction response in the sagittal plane for the control trial (4.9, 3.5 times larger for pelvis straps and pelvis + trunk support, respectively). The interaction response was not calculated between frontal and transverse planes since simulator motion in the frontal and transverse planes occurs

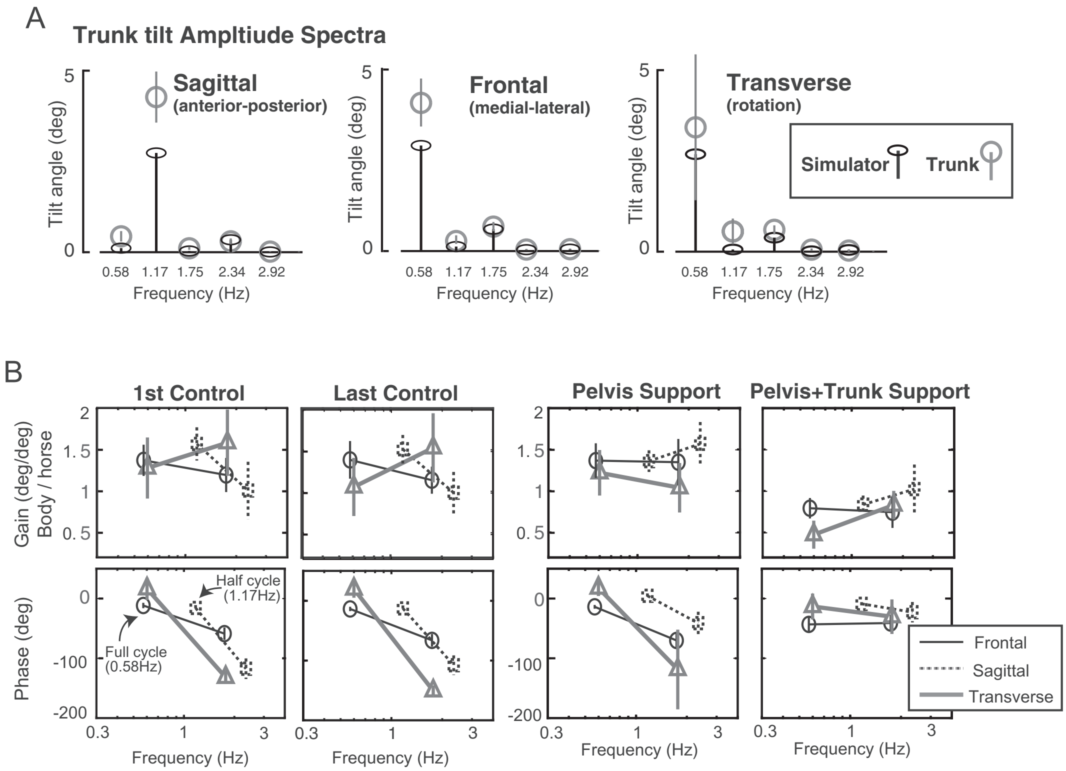

Fig. 5. A) Trunk amplitude spectra averaged across participants with 1 SD (grey) and simulator motion (black) for one control trial. To focus on the response to simulator motion (which was negligible at low frequencies), the Fourier transform was applied to the average cycle response of 1.712 s (corresponding to a fundamental frequency of ~0.585 Hz). B) Average participant gains and phases (with 95 percent confidence intervals) calculated for all trials from the relation between simulator stimulus and postural response at two frequencies where simulator motion was highest in each plane. A gain of 1 and phase of 0 indicates perfect alignment of the trunk to the simulator motion with no lead or lag in timing. A gain of less than one indicates lower sensitivity to the simulator motion and a negative phase indicates a lag in responsiveness to the simulator motion at a particular frequency. Frontal and transverse plane data is slightly offset from each other in frequency for visualization.

at the same frequencies.

The magnitude and timing of trunk main responses are described by pairing gains with phases for all trials in Fig. 5B, which displays the average participant and 95% confidence intervals at each frequency. Gains and phases were quite similar between the two control trials. In these control trials, gains were lower at higher frequency points for motion in the frontal (solid black lines) and sagittal (dotted lines) planes indicating that participants were less responsive to the simulator motion at higher stimulus frequencies in these planes. However, in the transverse plane (solid grey lines), there was a notable gain increase at higher frequency points, indicating an increased responsiveness to the simulator motion. Phases were lower at the higher stimulus frequencies, indicating more phase lag in responses at these frequencies.

The inclusion of pelvis support mainly impacted responses to the higher frequencies in three ways: 1) decreased the gain in the transverse plane (grey line) at 1.75 Hz, 2) increased gains in the frontal and sagittal planes at 1.75 Hz and 2.3 Hz, and 3) reduced the phase lag in the sagittal plane at 2.3 Hz. Finally, pelvis + trunk support trials clearly reduced gains across all frequencies and planes of motion and generally eliminated the large phase lag at higher frequencies.

3.3. Head responses in control condition

While the focus of this study is on trunk responses, head kinematics are described as a comparison in Fig. 6 for the control condition. There were several similarities between head and trunk motion. First, head responses generally mimicked the simulator motion (Fig. 6A). Second, head motion contained low frequency amplitude with clear main responses (Fig. 6C). Third, there was low interaction between planes (Fig. 6C). The ratio of the main response to interaction response was 5.9, 6.3, and 2.0 for the frontal, transverse, and sagittal planes, respectively. The lower ratio in the sagittal plane corresponds to the reduction in the main response in the sagittal plane (Fig. 6A, grey bars in RMS position and RMS velocity). Fourth, similar to the trunk, variability across repeated cycles and CV was lowest in the frontal plane and highest in the transverse plane (Fig. 6B), although this difference did not reach statistical significance ($p = 0.075$).

There were three notable differences between head and trunk motion. First, head motion was much more variable across repeated simulator cycles (Fig. 6B) and across participants (Fig. 6A – note the wider range in average head responses across participants

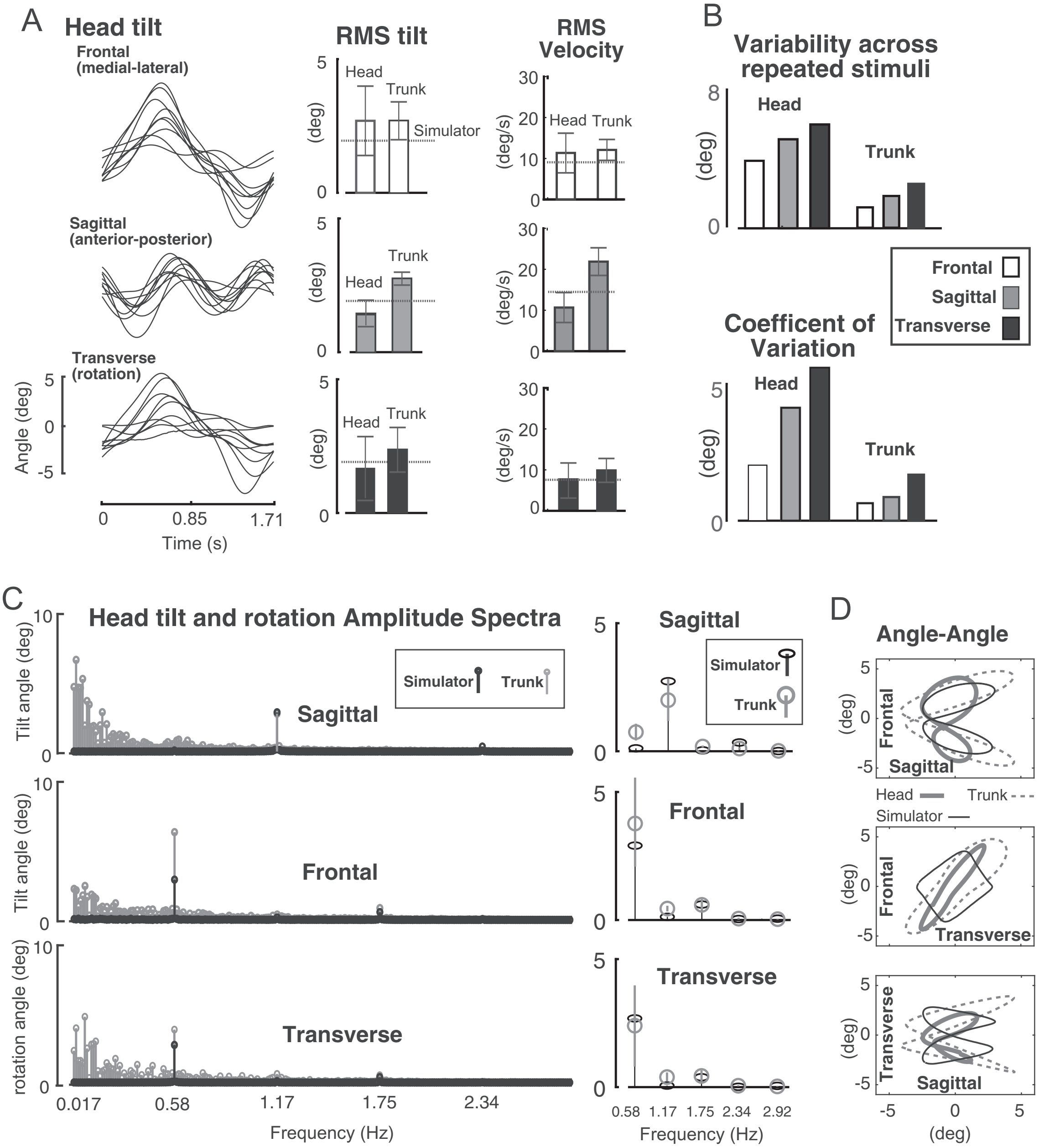

Fig. 6. Head kinematics for control condition. A) Head motion for all participants and average participant root mean square (RMS) position and RMS velocity with 1 SD for head with a comparison to trunk behavior B) Average subject variability across cycles and coefficient of variability (CV), see Methods for calculation details. C) Amplitude spectra for one representative participant using the entire 60 s trial in the Fourier transform to show low frequency head motion; average amplitude spectra across participants (1SD) at frequencies of simulator motion using the average cycle of 1.712 s. D) Angle-angle plot for head, trunk, and simulator averaged across participants.

corresponds to a larger error bars in RMS position Fig. 6B). CV of the head was significantly larger than CV of the trunk in the frontal ($p = 0.002$), sagittal ($p = 0.001$), and transverse ($p = 0.012$) planes. This variability mostly showed up across low frequencies (note the vertical scale in Fig. 6C is double Fig. 4). Second, RMS position and RMS velocity of the head was significantly lower than the trunk in the sagittal planes ($p < 0.001$), but not in the frontal or transverse planes (Fig. 6A, grey bars). Also, across planes of motion, there was a significant difference in head RMS position ($p = 0.01$), but not RMS velocity ($p = 0.073$). RMS position of head in the sagittal plane was lower than transverse or frontal. The lower motion in the sagittal plane is also evident in the angle-angle plots (Fig. 6D). These plots, in general, show a lower head response in the sagittal and transverse planes compared to the trunk and also revealed an asymmetry such that lower sagittal plane head motion was associated with rightward tilt in the frontal plane (top plot, Fig. 6D) and rightward rotation in the transverse plane (bottom plot, Fig. 6D). Third, the head responses had more phase lag in the sagittal and transverse planes at and above 1.17 Hz (data not shown).

Pelvis straps had minimal impact on head kinematics, but trunk support was associated with lower head motion (reduced RMS position and low frequency gains across all planes, data not shown). Thus, increasing stability near the low ribs translated to decreased head motion during the ride.

4. Discussion

The main goal of our study was to address fundamental and clinically relevant questions about kinematic postural responses to complex three-dimensional simulator motion. Each of the three questions posed for this study is addressed below.

4.1. What is the specificity of postural responses to the simulator riding motion?

There was a high degree of specificity in postural responses to simulator direction and frequency, suggesting that postural control could reasonably be understood through a linear lens during the experiments. Trunk postural responses and simulator motion visually matched each other closely in the time domain (Fig. 2), and the pattern of sagittal plane motion was very similar to anterior-posterior saddle pressures reported previously during hippotherapy (Fig. 1 in Janura et al., 2009). The angle-angle diagrams of trunk motion suggest a coordinated and symmetrical three-dimensional response. However, decomposing simulator and trunk motion in the frequency domain was particularly helpful to address the question of specificity. Specifically, postural responses in the sagittal plane occurred with a notable amplitude at exactly the same two frequencies that the sagittal plane simulator tilt occurred (main response). While there was some sagittal plane response at the frequencies where frontal and transverse plane simulator motion occurred (interaction response), this was minimal (Figs. 4 and 5). The ratio of main response to interaction response was on average 6 to 9 times larger. The same general finding was true for frontal and transverse planes though we cannot completely rule out an interaction between frontal and transverse planes since simulator motion in those two planes occurred at the same frequencies.

Given the multi-articulate trunk musculature and that muscle fiber orientation is not along anatomical lines, the high linearity in responses may seem surprising. Synergistic activation could account for the stereotypical and directionally specific responses (Torres-Oviedo & Ting, 2007). Also, these linear responses are consistent with previous posture studies that elicited sway using a distinct mode of stimuli (surface tilt, visual tilt, or galvanic vestibular stimulation) in either the sagittal or frontal plane and found responses at discrete frequencies corresponding to the stimuli in that plane of motion (Cenciarini & Peterka, 2006; Kiemel et al., 2008; Goodworth & Peterka, 2009; Peterka, 2002; Mergner et al., 2005; Wu, Duncan, Saavedra, & Goodworth, 2016). In response to either frontal or sagittal plane stimuli, the typical pattern of gains and phases across frequency involve gain increases from about 0.01 to 0.8 Hz followed by gain decreases at frequencies above 1 Hz (Goodworth & Peterka, 2010; Peterka, 2002). Phases typically show more lag with increasing frequency. Modeling has indicated that low – ($<$ 0.2) and mid-frequency ($<$ 2 Hz) gains and phases are highly influenced by sensory feedback, whereas higher frequencies ($>$ 2 Hz) are more influenced by reflexive processes and intrinsic biomechanics (Goodworth & Peterka, 2009). Body inertia and neural time delays can also contribute to the observed gain decreases and phase lags with increasing frequency. In the current study, we found gain and phase decreases with frequencies above 1 Hz in the frontal and sagittal planes (Fig. 5), suggesting that control processes (neural gains, sensory reliance, contribution of intrinsic musculoskeletal stiffness and damping, and neural time delays) identified in studies with a single plane stimulus may be applicable to responses during a more complex three-dimensional perturbation.

The presence of low frequency body motion (trunk in Fig. 4, head in Fig. 6) is similar to that found in spontaneous sway (unperturbed standing and sitting) (Genthon, Vuillerme , Monnet , Petit, & Rougier, 2007; Goodworth & Peterka, 2018), which is characterized by low frequency sway that significantly lowers in power proportional to 1/frequency (Boonstra, Schouten, & van der Kooij, 2013) or a low passed filtering of 1/frequency (Goodworth & Peterka, 2018; van der Kooij & Peterka, 2011). At the lowest frequencies of simulator motion (0.58 Hz and 1.17 Hz), spontaneous trunk tilt amplitude is expected to be quite small, dropping by an order of magnitude from 0.04 Hz to 1 Hz (Goodworth & Peterka, 2018). To more definitively understand the relation between the simulator and low frequency trunk and head motion, a future study would need to include trials where participants sit on the simulator while it is not moving.

While our data suggest a high degree of linearity across planes of motion and frequency of perturbation, it does not imply that motion in one plane has no impact on the amplitude of postural responses in other planes. Previous studies have demonstrated an inter-modal sensory reweighting where one mode of stimuli can change responsiveness to another mode (Hwang et al., 2014; Cenciarini & Peterka, 2006; Mergner et al., 2005). For example, people increase their reliance on vestibular cues when responding to large surface tilts, and this increased reliance heightens responsiveness to galvanic vestibular stimulation (Cenciarini & Peterka, 2006). However, even in these inter-modal studies, responses to stimuli primarily occur within the plane of motion that the stimulus was delivered.

We found important differences in responsiveness across planes and with head control. In the transverse plane, trunk gains increased with increasing frequency. One explanation for this difference is that trunk rotational inertia and other biomechanical characteristics are quite different in the transverse plane than the other planes. Also, gravitational forces do not affect transverse plane rotations as they do for trunk tilt in the frontal and sagittal planes. Head control also demonstrated interesting differences across planes, with motion considerably diminished in the sagittal plane and (to a lesser extent) transverse plane compared to the frontal plane (Fig. 6). The difference may have arisen because head orientation has less impact on the upper body center of mass (due to its relatively lower mass) than the trunk, and it can be decoupled from the trunk posture system. The lower head motion resulted in more stable head orientation in the transverse and sagittal planes which could be used to help stabilize gaze, similar to findings in previous studies (Buchanan & Horak, 2001; Pozzo, Levik, & Berthoz, 1995).

Finally, our interpretation of postural responses to simulator motion was primarily based on simulator tilt and rotation. However, simulator motion also includes small translations. Translations along the horizontal plane could influence postural responses in the sagittal and frontal planes (Buchanan & Horak, 2001; Preuss & Fung, 2008), but would not be expected to have a direct impact on transverse plane motion. Peak translation characteristics of the simulator in the "x" direction (affecting sagittal plane responses) were 0.014 m, 0.1 m/s, and 0.85 m/s^2; and in the "y" direction (affecting frontal plane responses) were 0.006 m, 0.035 m/s, and 0.35 m/s^2. By comparison, a previous study elicited trunk postural responses using translation velocities of 0.45 m/s, which are about 4.5 and 13 times higher than the simulator in the "x" and "y" direction, respectively (Preuss & Fung, 2008). To further explore the influence of translations, we used a parameterized feedback model of trunk postural control, based on Goodworth and Peterka (2009), to simulate trunk responses with and without the surface translations (Matlab Simulink, The Mathworks, MA, USA). The model suggested that trunk tilt amplitudes were increased by the translations by approximately 12.5%, 3.5%, and 0% in the sagittal, frontal, and transverse planes, respectively. Translations in the "x" direction primarily occurred with the same timing and frequency as sagittal plane simulator tilts. Translations in the "y" directions similarly corresponded with frontal plane simulator tilts. Thus, the main conclusions about specificity of direction and frequency hold with this combination of tilt and translation from the simulator motion.

4.2. Which plane of motion is associated with the most variability?

Variability across both repeated cycles and participants was greatest in the transverse plane. In contrast, variability in the frontal plane was quite low (Figs. 2 and 3). This result provides insight into the behavioral goal. Controlling motion in the transverse plane is likely least important for the posture task because the effect of gravity is less than the other planes. Thus, its variability is free to be higher. But, why is variability higher in the sagittal compared to frontal plane? One explanation is the different threat of injury between planes. The simulator (and live horses) offer very little margin of error in the frontal plane, where sliding off the saddle toward either side is a risk. In the sagittal plane, a "fall" would look like a participant falling onto the front or back of the seat (or saddle). In the frontal plane, a "fall" would look like a participant literally falling off the horse. Previous posture studies have shown significant differences in muscle activations and behavior as function of postural threat, such as when participants' posture is measured after being lifted high above the ground (Carpenter, Frank, Silcher, & Peysar, 2001).

Similar to riding the simulator, lateral falls while walking can be particularly dangerous (Mayhew et al., 2005). Based upon previous walking model studies, humans are passively stable in the sagittal plane while inherently unstable in the lateral direction (Kuo, 1999). Experimental results in which visual and physical perturbations were applied in the sagittal and frontal planes while people walked on a treadmill also indicted a larger lateral instability (McAndrew, Wilken, & Dingwell, 2011). Also, pelvis variability during walking is generally lower in the frontal plane compared to the sagittal or transverse planes based on intra-participant coefficient of variation values (Vogt, Portscher, Brettmann, Pfeifer, & Banzer, 2003) and inter-participant standard deviation values (Whittle & Levine, 1999), similar to variability across planes in the current study. Therefore, in addition to similarity in pelvis kinematics between horse riding and walking (Garner & Rigby, 2015), there is also similarity in control processes across planes of motion between horse riding and walking; and these combined factors might be responsible for improved gait mechanics following horse riding in those with impaired walking.

Another interpretation of the variability in the current study is that motion in each plane is controlled through independent neural systems, each with their own inherent noise (Faisal et al., 2009). Evidence for this idea comes from a number of gait studies that showed sagittal plane motion (step length) can be adapted independent of rotation in the transverse plane (foot and pelvis rotation needed for turning) during treadmill walking (Reisman, Block, & Bastian, 2005). The reverse is observed after stepping on a rotating disk (Earhart et al., 2001); and variability in the transverse plane is significantly higher than in the sagittal plane during circular walking (Goodworth et al., 2012). Regardless of the cause, there is evidence that practice in hippotherapy can reduce variability (Janura et al., 2009).

Finally, head variability was notably higher than variability of the trunk (Fig. 6). Again, this could have arisen because head orientation (being relatively lower mass) has less impact on the overall upper body center of mass than the trunk. However, while participants were instructed to face forward, there were a few occurrences of participants momentarily turning his/her head, which would of course slightly increase measures of variability.

4.3. How do pelvis straps and trunk support impact postural responses?

The most obvious effect of the mechanical modifications was in the pelvis + trunk support condition. In this condition, trunk responses were significantly lowered in RMS position and RMS velocity (especially in the transverse plane), lower in gains across all frequencies, and phase shifted closer to zero (more synced in time with the simulator motion) across all planes (Fig. 5). Trunk support therefore causes a fundamental shift in the timing of responses (not just downscaling of trunk amplitude). The trunk support raised the base of support to the middle of the trunk and reduced the degrees of freedom in the spine that needed to be controlled against gravity.

In contrast to pelvis + trunk support condition, the pelvis straps alone had minimal effect on RMS position, RMS velocity, and gains and phases where motion was largest (low frequency in each plane in Fig. 5). This finding may be explained by the fact that healthy participants were already able to maintain vertical pelvis alignment at the lower frequencies. Interestingly, however, the pelvis straps had a clear and significant influence at higher frequencies by increasing trunk tilt in the sagittal plane, reducing the trunk phase lag in the sagittal plane, and reducing transverse plane trunk rotation. The pelvis straps were designed to fix posterior pelvis tilt found in populations with poor trunk motor control. This modification targets the sagittal plane by orienting the pelvis more perpendicular to the surface the participant is sitting upon, so it is not surprising the sagittal plane trunk responses were influenced most. Because the sagittal plane trunk tilt calculation used the midpoint of the pelvis as a reference, it is reasonable to question how much this reference point impacted trunk tilt. After re-analyzing trunk data in absolute displacements (without reference to the pelvis), we found similar effects of the pelvis strap on high frequency changes in trunk motion to those reported in Fig. 5. Regarding the transverse plane, we note that the straps were tight and therefore had the potential to impact all three planes of motion. We can only speculate that the reduction in transverse plane trunk motion happened because the straps made the pelvis more stable on the moving surface (transverse plane pelvis rotation was about 25% lower at the higher frequency in each plane with the pelvis straps compared to control conditions).

4.4. Clinical and research implications

Hippotherapy and riding a mechanical horse has documented benefit in several populations for both gait and balance (Bronson, 2010; Dewar et al., 2015; (Lechner et al., 2007). With practice, postural responses can improve (reduction in magnitude) (Janura et al., 2009), and the size of trunk muscles can increase (Park & You, 2018). The horse motion evokes patterns in the pelvis similar to walking over ground (Garner & Rigby, 2015), and the massed repetition requires patients to generate and practice many postural responses. While the following section focuses on clinical implications for hippotherapy and simulated hippotherapy, it should be noted that many implications also apply to general perturbation training for balance control, which is growing in popularity due to its documented effectiveness and theoretical underpinning (Goodworth, Perrone, Pillsbury, & Yargeau, 2015; Horak, Henry, & Shumway-Cook, 1997; Mansfield, Wong, Bryce, Knorr, & Patterson, 2015).

In the current study, the high specificity of postural responses imply that one could strategically select or adjust the pattern of stimuli in each plane to initiate specific postural responses. For example, it is well known that different horses exhibit different three-dimensional patterns (Jámbor, Bokor, & Stefler, 2009), and horses are typically categorized and selected for hippotherapy based on walking patterns and the desired outcomes for specific patients. With the advancement of sensor technology, it should be more accessible to quantify horse kinematics and thus provide more specificity for therapy goals. Quantification of horse motion could also be used to avoid certain motions, such as high velocity motion patterns sometimes associated with spasticity (Shumway-Cook & Woollacott, 2017).

If any patient exhibits coupling between planes of motion significantly higher than those observed in the present study, this could be indicative of abnormal associative movements or abnormal synergies (Shumway-Cook & Woollacott, 2017). Similarly, abnormal sensitivity in one plane suggests an abnormal response. For example, high axial tone (characteristic of PD), would likely show reduced sensitivity to translational motion of the horse (Wright, Gurfinkel, Nutt, Horak, & Cordo, 2007). Abnormal visual and vestibular processing would likely show increased frontal and sagittal tilt motion because visual and vestibular feedback help stabilize the trunk in response to surface tilts (Peterka, 2002; Goodworth & Peterka, 2010). As healthy participants demonstrated significant damping of head velocity in the sagittal plane, this may also be used as a benchmark for comparison in populations with poor head control (Saavedra & Woollacott, 2015).

Finally, the addition of a pelvis strap and/or trunk support has been shown to improve posture alignment and responsiveness in populations with underdeveloped postural control or motor impairments (Butler et al., 2010; Rachwani et al., 2015; Saavedra & Woollacott, 2015; Goodworth et al., 2017). Given the comfort in healthy participants and similarity between pelvis straps and control trials, we anticipate the straps will be comfortable in most patient populations and should help normalize postural responses in those who need straps for proper control. In contrast, the trunk support significantly changed the amplitude and timing of postural responses, and thus would not be expected to make responses similar to free sitting healthy participants. Nevertheless, trunk support is required for upright postural control in some populations (Butler et al., 2010; Saavedra & Woollacott, 2015), and our results supported the idea that trunk support makes the posture task easier, as evidenced by lower trunk and head responses.

4.5. Limitations

The currently study only addressed one speed and amplitude of simulator motion, but to understand adaptations, one could investigate multiple speeds and amplitudes, as both stimulus speed and amplitude can evoke changes in the postural control system (Mergner et al., 2005; Peterka, 2002). Only one type of trunk support was offered to participants, and this device altered the timing of trunk responses. One could conceive of multiple designs, each having a unique stabilizing effect and impact on trunk responses, which would have value to investigate in the future. Finally, all participants sat on the simulator without a saddle or stirrups, but it has been shown that different saddle configurations can influence the force transmission to the rider (Ribeirom, 2018).

5. Conclusion

Nine healthy individuals responded to complex three-dimensional simulator motion. There was high specificity between simulator motion and trunk responses across planes of motion and frequencies. The overall magnitude of trunk responses was similar across planes, though differences were found at higher frequencies. Head responses significantly differed across planes (smallest in the sagittal plane). Variability across repeated cycles and participants was highest in the transverse plane motion and lowest in the frontal plane, suggesting that frontal plane responses were highly controlled. Finally, the inclusion of pelvis support had minimal influence over low frequencies of trunk responses but altered higher frequencies, especially in the sagittal plane; and the inclusion of trunk support significantly reduced motion across all planes and frequencies making trunk responses more synchronized with the simulator motion. These results provide foundational knowledge about postural responses to complex stimuli and provide a baseline for comparison to pathological populations.

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Conflict of interest

Dr. Brian Garner is the inventor and developer of the mechanical horse-riding simulator that was used in this study. In consultation with the technology transfer and business acceleration resources of Baylor University, Dr. Garner has founded a company, Chariot Innovations, Inc., with the mission of making the mechanical horse and similar technologies available in the broader marketplace.

For the remaining authors, no conflicts of interest, financial or otherwise, are declared.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.humov.2018.11.007.

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