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The effect of repeated shocks on the low back during horse riding

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ABSTRACT

Repeated shocks, such as those occurring during horse riding, may result in a risk of overloading the low back. This study investigated accelerations and angular changes in the low back during horse riding, using Inertial measurement units (IMUs) on the pelvis, and L4, L1, and T11 spine levels during 30 min of walking and 10 min of cantering in twelve female participants. The root mean squared (RMS) linear acceleration of each IMU and the transmission (signal transfer) between each pair of IMUs were calculated from the measured accelerations. Additionally, angular motions between IMU's were quantified. The RMS of vertical pelvis accelerations was overall higher for cantering than for walking, with a peak value of 5.52 vs. 0.85 m/s^2 at 1.8 Hz. Transmission of accelerations was 1 or slightly above 1 at 4 Hz, indicating pelvis accelerations to be passed on to the low back equally or somewhat amplified. Above 4 Hz, accelerations were damped, increasingly so with increasing frequency. Damping was the largest between the lowest segments. Rapid (<0.1 s) and relatively large (up to 30°) angular changes were found between the Pelvis and L4 sensors. High-frequency angular movements indicate involuntary movements, likely as a result of impact. Potentially, angular changes and spine compression resulting from the accelerations during shock loading in horse riding could exceed injury thresholds.

1. Introduction

Low back pain (LBP) has a relatively high prevalence in elite equestrians, compared to the general population. Literature shows that the one-year prevalence of LBP is 61–74 % for elite equestrians, compared to 22–65 % in the general population (Duarte et al., 2024; Hoy et al., 2012).

Spinal injuries constitute a possible cause of LBP, as shown by the association between mechanical loading and LBP incidence (Coenen et al., 2014; Griffith et al., 2012; Norman et al., 1998). Excessive vibration and repeated shocks, common in activities such as horse riding, off-road cycling or driving, and sailing in high-speed boats, are risk factors for LBP (Barrero et al., 2019; Pope et al., 1996; Waters et al., 2007). Horse riding is the activity for which earlier research has provided most insight into the relationship between exposure duration and intensity and risk of LBP (Ferrante et al., 2021; Kim et al., 2020; Kraft et al., 2009; Lewis and Kennerley, 2017). While riding a horse in walking gait was proven useful as therapy for people with existing LBP (Kim et al. 2020,), more strenuous horse riding activities, such as dressage, cantering and showjumping, may result in LBP after training periods of five to six hours per week (Ferrante et al., 2021; Kim et al., 2020; Kraft et al., 2009). Previous research, mainly questionnaire-based, lacks detailed insights into spinal shock transmission and potential excessive spine bending or buckling (Meakin et al., 1997; Preuss and Fung, 2005).

Axial spine loading, as caused by exposure to shocks and perturbations, can cause injury along two mechanisms. First, compressive loading can cause endplate fractures in vertebrae (M. (Adams and Roughley, 2002; Brinckmann et al., 1988, 1989), which are associated with LBP history (Wang et al., 2012). Repeated pressure on the intervertebral disc causes it to lose fluid, changing its dynamical properties and the pressure distribution over the endplate (Adams et al., 1996; Alkalay et al., 2015; Van Dieën et al., 2001). Therefore, repeated axial shocks of substantial magnitudes are likely to cause spinal injury though repeated spinal compression (Brinckmann et al., 1988). Second, sudden axial loads on the trunk may damage soft tissues like spinal ligaments due to bending or buckling of the spine (Preuss and Fung, 2005). Buckling may occur due to the spine's inherent instability, combined with sudden and large axial forces, causing it to bend relative to its neutral position (McGill and Cholewicki, 2001; Preuss and Fung, 2005). In an upright posture, spine stability is primarily achieved through

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muscle co-activation (Cholewicki and Mcgill, 1996), increasing muscle stiffness in anticipation of a perturbation (Solomonow et al., 1999). Spine stability is also achieved through reflexes (van Dieën et al., 2018; van Drunen et al., 2013). Fatigue and creep loading of connective tissues can decrease these anticipatory and reactive capacities, increasing buckling risk (Solomonow, 2012). Additionally, sudden large lumbar flexion combined with dynamic shear loading of the spine can result in flexion-distraction injuries, including both ligament ruptures and disruption of ligaments from the spine (Neumann et al., 1995).

We aimed to provide insight into effects of shock loads on the lumbar spine of equestrians during horse riding in walk and canter. First, as damping and amplification of acceleration may reduce and increase compression peaks, respectively, we evaluated the magnitude and transmission through the low back of vertical linear accelerations. In line with earlier studies, we expected that amplification would occur at 4 HZ, and damping at higher and lower frequencies (Pope et al., 1996). Second, we evaluated the magnitude and velocity of local angular changes in the low back. We hypothesized both acceleration peaks and rapid angular changes to be larger in cantering than in walking.

2. Methods

2.1. Participants

Twelve female participants (mean (SD) age: 30 (10) years; horse riding experience: 20 (10) years) were included in this study. Equestrians were not allowed to participate if they had experienced back injuries or back pain that led to modifications in their riding behaviour in the last six months.

This study was approved by the ethics committee of the Faculty of Behavioral and Movement Sciences of the Vrije Universiteit Amsterdam (VCWE-2023–076) and all participants signed informed consent before participating.

2.2. Set up

3D linear accelerations, angular velocities, and magnetometer data were gathered using inertial measurement units (IMUs; Movella DOT sensors, Enschede, the Netherlands). One IMU was attached to the back of the horse's saddle, and four IMUs were attached over the equestrian's spine, at the height of the sacrum, and the L4, L1, and T11 vertebrae. IMUs were attached using double-sided tape and stretch tape to ensure minimal movement relative to the skin. Sampling frequency was set to 120 Hz and a dynamic filter was used to stabilise the heading of the IMU by assuming magnetic disturbances to have a very short duration due to fast movements (Alcala et al., 2023). Quaternions (based on the Movella onboard fusion algorithm), angular velocity, linear acceleration, and magnetic field data were extracted (Movella DOT Data Exporter, 2022). In the remainder of this paper, *acceleration* will be used to refer to linear accelerations, unless stated otherwise.

2.3. Synchronisation

The IMUs were synchronised via onboard synchronisation and mechanical synchronisation before attachment to the participant. IMUs in a rigid case were tapped on a hard surface several times to generate sharp acceleration peaks, which were aligned during data processing to correct synchronisation if needed.

2.4. Calibration

After attaching the IMUs, calibration measurements were conducted to serve as posture reference and to align the IMUs' frames. The equestrian stood still for five seconds and walked in a straight line of approximately 10 m. Realignment of IMUs was achieved by estimating the axis directions from the mean accelerations during walking. The best harmonic ratios product was used to recalculate the horizontal axes and associated rotation matrices. These rotation matrices were used to rotate the raw accelerations and angular velocities in order to align all IMUs (x forward, y left, z up).

2.5. Trials

The first measurement consisted of the equestrian riding in walking gait for 30 min. For the second measurement, the equestrian was instructed to canter for 10 min. If they were unable to do so, because of their condition or that of their horse, they were allowed to split the cantering period in two or three episodes. Equestrians were allowed to ride in circles and change directions, as long as the horse did not walk backwards or sidewards.

2.6. Data analysis

A dedicated MATLAB (MathWorks Inc, version 2023b) script was employed to preprocess the data using the aforementioned synchronisation and calibration steps. The raw data were then visually checked to rule out and remove major inconsistencies and errors.

Using the quaternion data of the IMU's, a semi-world-fixed frame was constructed, in which the vertically aligned IMU axis is fixed to a global world-fixed vertical axis, around which the local horizontal axes rotate, keeping the x-axis in the walking direction. This was done by rotating the local data to the global frame using quaternions obtained directly from the sensor output according to the built-in algorithm of the manufacturer, and subsequently multiplying the global data with the transpose of a rotation matrix describing the rotation around the global vertical axis. The combined frame prevents irregular distribution of movements over the horizontal axes due to circular riding. The calibrated data were used to calculate the orientation of the IMUs with respect to the semi-world fixed frame during walking and cantering. These steps were validated previously (Smit et al., 2025).

The orientation of the IMUs in the above mentioned combined frame, was calculated using Euler decomposition (zyx, starting with axial rotation due to circular riding) of the quaternions. Angles of IMUs relative to each other were calculated by taking the Euler decomposition (yxz, relative axial rotation between IMUs should be close to zero) of the quaternions expressing the orientation of the proximal IMU with respect to the distal IMU. Angular accelerations were calculated through differentiation of the angular velocity signals. Linear velocity and displacement were calculated by numerically integrating the accelerations. The effects of drift were removed using a second order high-pass Butterworth filter with a cutoff frequency of 0.1 Hz. From accelerations in the combined frame, root mean squared acceleration (RMS) of IMUs, and transmissions of IMU accelerations relative to the pelvis IMU were calculated. Transmission and RMS results were calculated for one-third octave frequency bands with upper and lower bounds (Eq. (1), and centre frequencies (Eq. (2):

$$\text{FreqBorder} = 2^{\left[-1:\frac{1}{3}:5\right]} \tag{1}$$

$$\text{FreqCenter} = 2^{\left[-\frac{5}{6}:\frac{1}{3}:4\frac{5}{6}\right]} \tag{2}$$

For transmission, separation into frequency bands was done using the transmission and frequency output of the tfestimate function in MATLAB (Kingma and van Dieën, 2009). For the RMS, the frequency separation was done through bandpass bidirectional filtering (3rd order) of acceleration data using the frequency border values from Eq. (1) as cut-off frequencies.

To quantify the magnitude and duration of relative flexion/extension angular changes between the pelvis and each other IMU, the magnitude of sections of angular change and their corresponding duration were determined. The MATLAB function peakfind was employed to find peaks in the relative angular velocity of at least 60 deg/s, of which the timepoints were selected in the relative angle between IMUs (Fig. 1). Sections of the relative angle signal were then selected by finding the closest changes in direction (e.g. flexion to extension). Multiples of the same interval were removed.

For two participants, the L4 IMU failed to record the trials. For measures making use of data from this sensor, these two participants were excluded.

2.7. Statistics

A Statistical Parametric Mapping (SPM) repeated measures one-way ANOVA was conducted to test differences between walking vs. cantering for Pelvis-T11 transmission. An SPM repeated measures two-way ANOVA was conducted test for effects of gait pattern (walking vs. cantering) and sensor level (pelvis vs. T11) and their interaction. A significance level of $p < 0.05$ was used.

3. Results

Vertical acceleration peaks were smoother and somewhat lower for the T11 than for the pelvis IMU for both walking and cantering (Fig. 2E). The majority of vertical acceleration peaks, excluding g, was below 5 m/s^2 for walking (Figs. 3A and 3C) and between 20 and 40 m/s^2 for cantering (Figs. 3B and 3D). The peak RMS was 0.85 (SD = 0.73) for walking, and 5.29 (SD = 1.08) for cantering, both at 1.8 Hz (Fig. 4A). Flexion-extension angles showed a difference in riding technique between participants for walking (Fig. 2B and S1), but were more similar between participants during cantering (Fig. 2B and S2), where high frequency components of angular motions were strongly reduced from the pelvis to T11 IMUs (Figs. 2B, 2D, and 2F). Vertical accelerations were much larger than the accelerations along other axes (Figs. S3-S4).

The reduction of high-frequency content over the lumbar spine is also clearly visible in the RMS and transmission per frequency band for the vertical accelerations (Fig. 4). Specifically, the transmission between the pelvis and T11 IMUs for frequencies higher than 3.56 Hz was less than one, indicating damping. The RMS acceleration shows large peaks at 1.78 and 3.56 Hz (Fig. 4A), which are caused by the periodic movement of the body depending on the stride frequency of the horse. Transmissions suggest a slight amplification of accelerations between pelvis and T11 IMUs at frequencies below 5.7 and 4.5 Hz for walking and cantering respectively (Fig. 4B).

The damping and amplification of the accelerations and angular velocities happened gradually between the pelvis and T11, with the largest change between the lowest three IMUs, as illustrated by example data (Figs. S5-S6). While RMS accelerations (Fig. 4A) show major differences between walking and cantering, transmissions across frequency bands (Fig. 4B) show a similar pattern between gait types. The F statistics can be found in Figs. S7-S8.

Maximum relative angular changes of Pel-T11 and Pel-L4 were found to be approximately 30° for walking and 40° for cantering. Relative angular changes between the Pelvis and L4 IMUs and between Pelvis and T11 IMUs were found to exceed 20° within 0.1 s numerous times in 10 min of cantering (Table 1, Fig. 5), with six out of ten participants reaching this limit at least one time. This was not the case for walking (Fig. S9). Relative L4-L1 and L1-T11 angular changes were considerably smaller than between pelvis and L4 IMUs (Table 1, Fig. 6 and S10).

4. Discussion

An obvious explanation for cantering posing a higher risk of LBP than

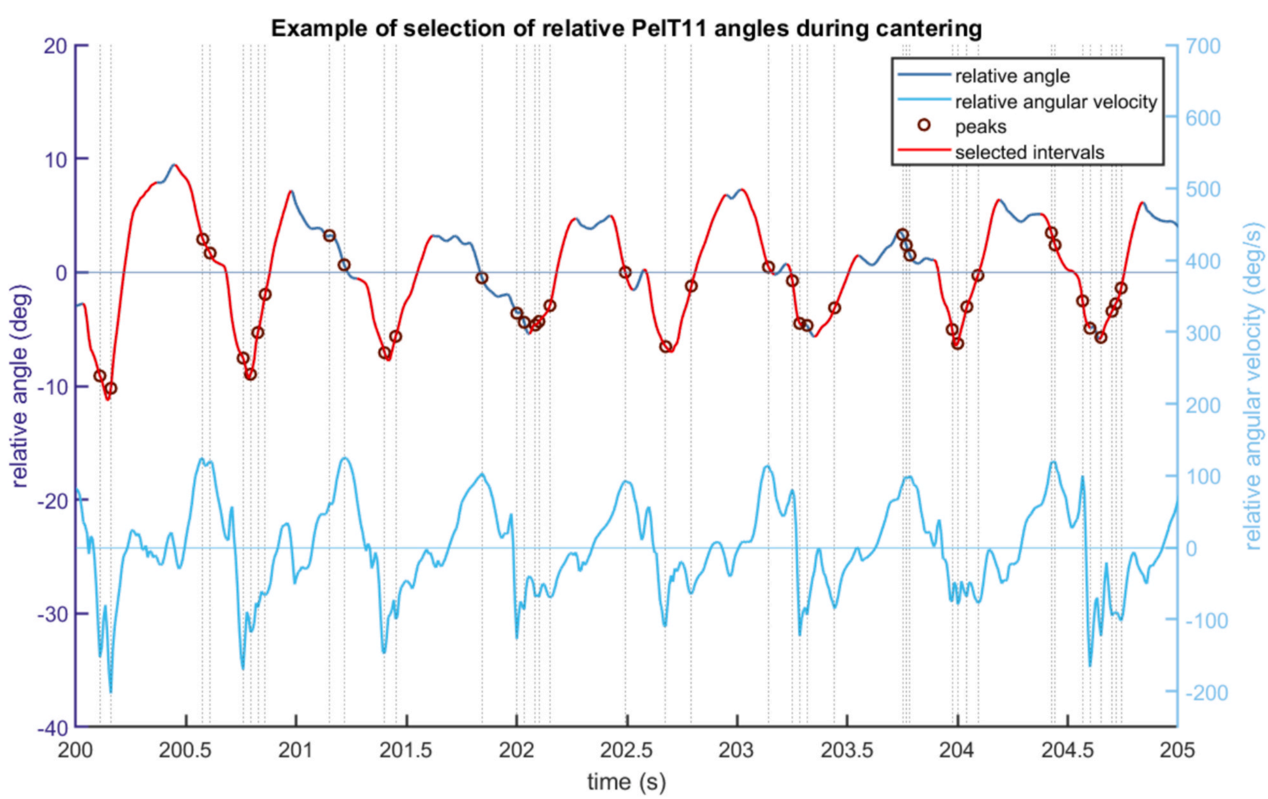

Fig. 1. Example of the selection process of relative flexion (positive)/extension (negative) angular change intervals. Relative angle between the pelvis and T11 vertebra (dark blue), with corresponding angular velocity (light blue). Selected peaks from angular velocity (>60 deg/s) are indicated with the black circles and vertical lines. Selected intervals (>0,04 s) are given in red. Redundant overlapping intervals were removed. For example, the positive change in relative angle between 204.5 and 205 s includes 4 points for which a peak occurred in the relative angular velocity. For each of these points, the interval was selected by finding a change in direction in relative angle. These four intervals are equal and only one of them was used for further analysis. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

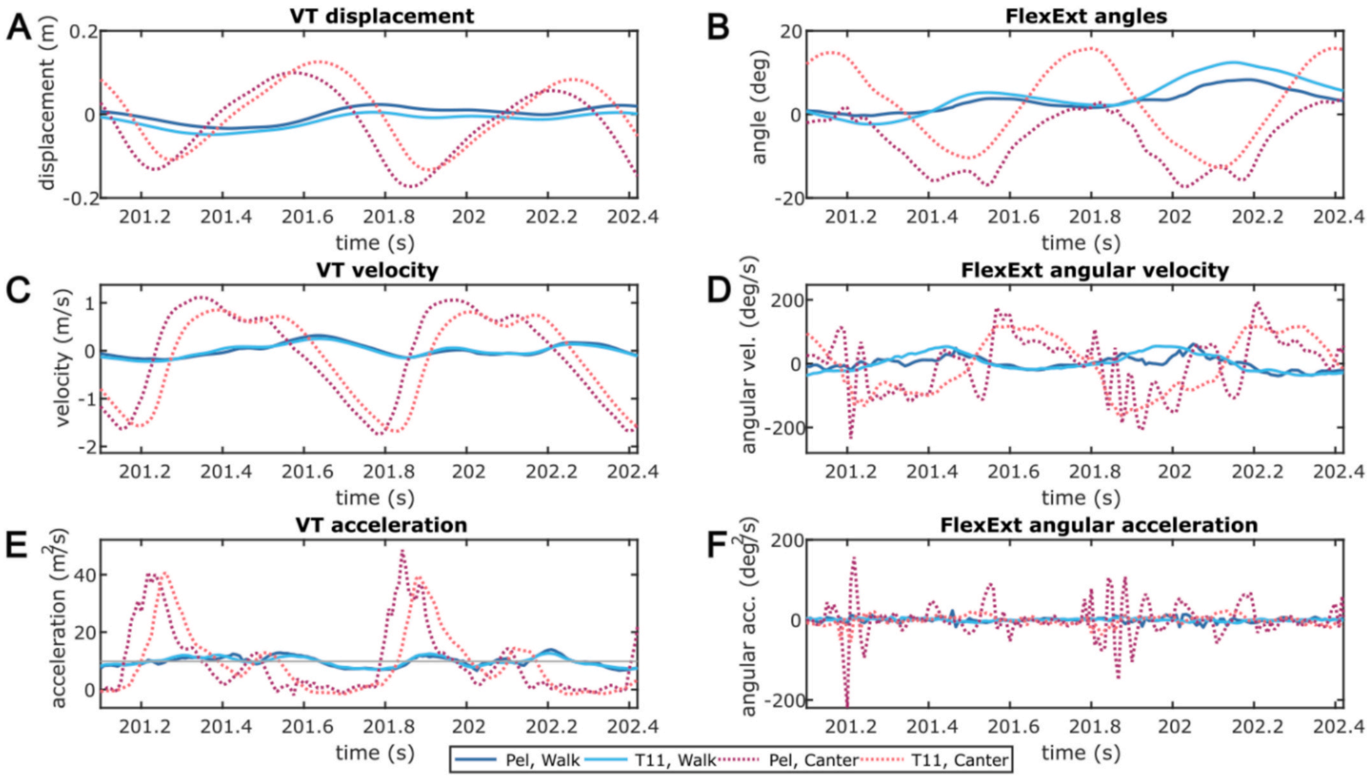

Fig. 2. Sample of linear vertical (A, C, E) and angular flexion (positive)-extension (negative) (B, D, F) movement of the pelvis and T11 of one participant (Sub2) during walking and cantering. A, B: displacement, C, D: velocity, E, F: acceleration. Note that linear acceleration included gravitational acceleration (g, black line). Note that small but rapid changes in angles, associated with large angular velocities and angular accelerations, are seen in the pelvis IMU but not in the T11 IMU.

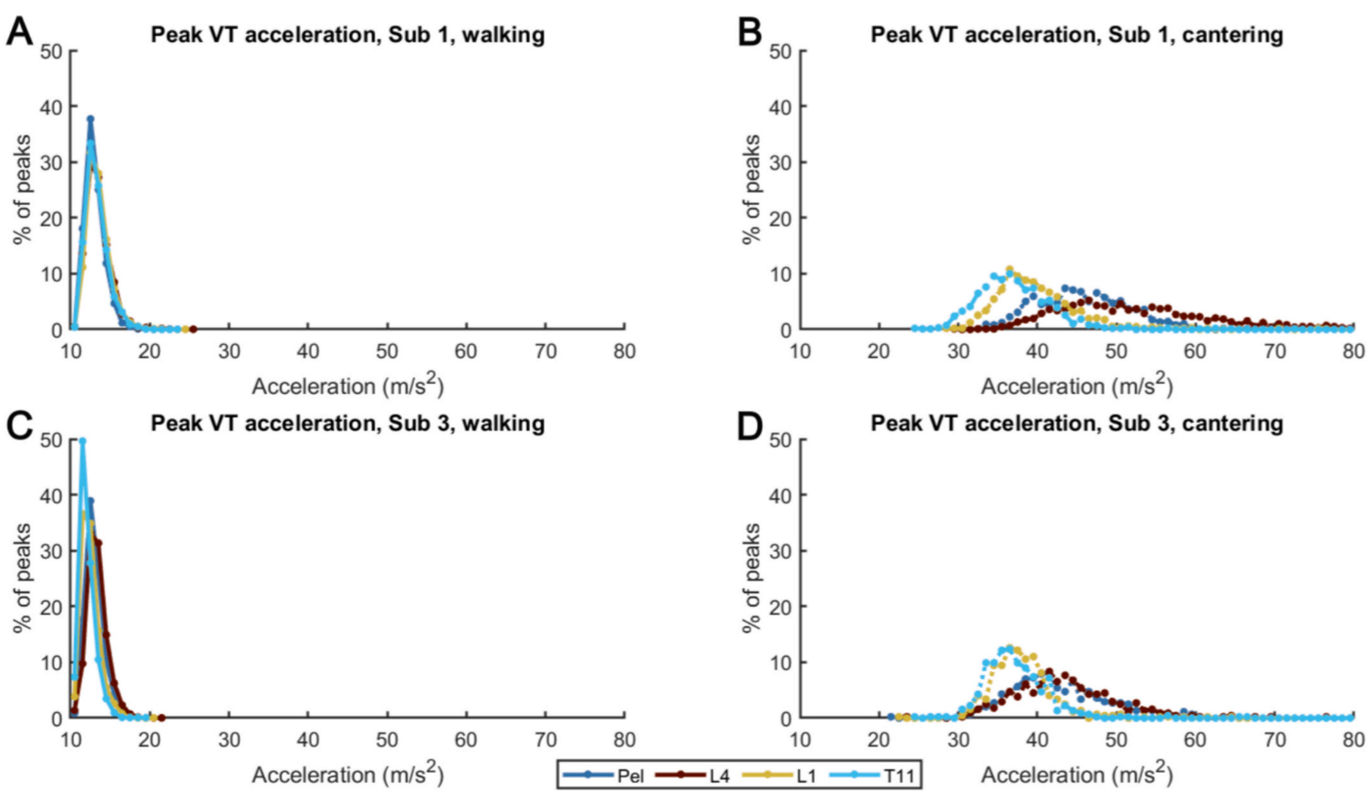

Fig. 3. Relative number of acceleration peaks of at least 10 m/s^2 (gravitational acceleration, g) for two participant and each measurement site for walking (A, C) and cantering (B, D).

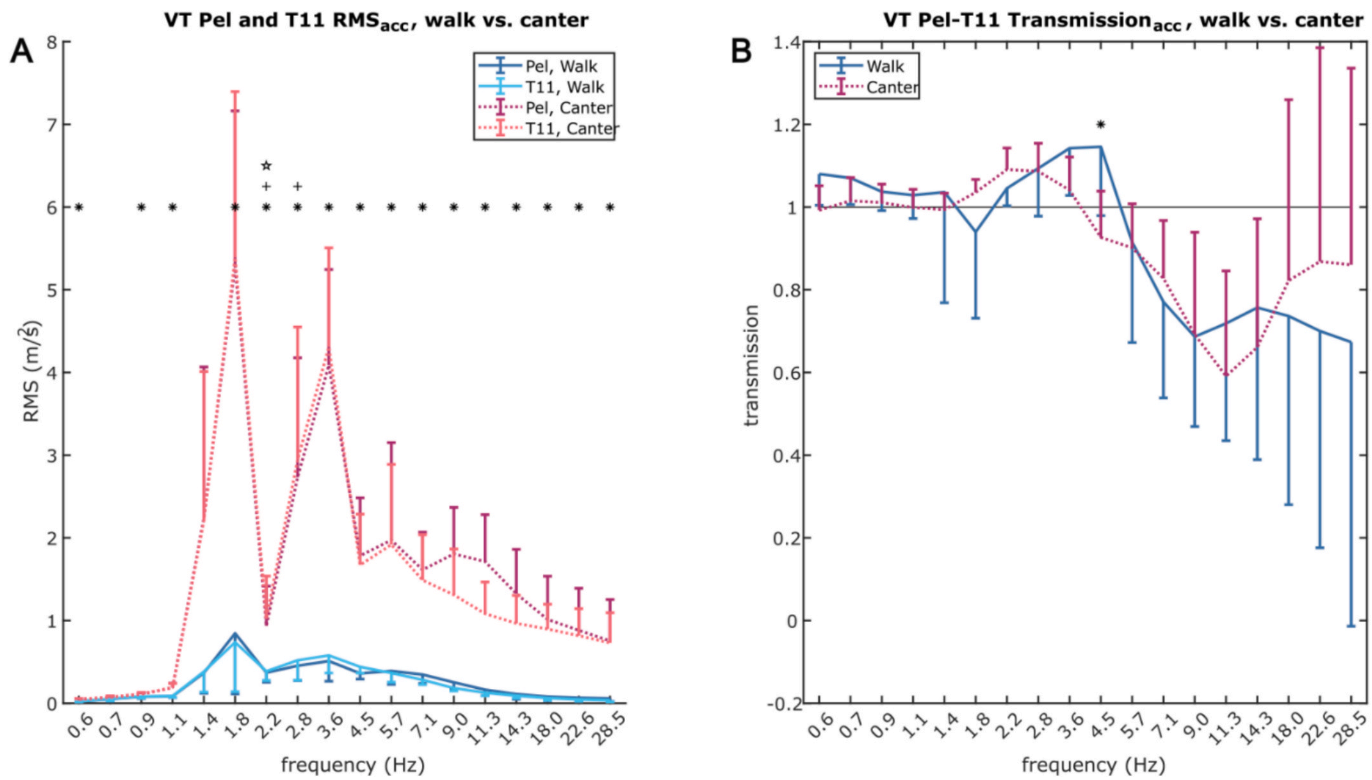

Fig. 4. Mean (and SD) root mean square (A) and transmission (B) of vertical accelerations of the pelvis and T11 sensor during walking and cantering per one third octave frequency band. The center frequencies are used to indicate each frequency band. The * indicates significant (<0.05) difference, for RMS between RMS walk and RMS canter, for transmission between Pelvis-T11 transmission during walk and canter. The + indicates significant (<0.05) difference, between Pelvis RMS and T11 RMS. The ☆ indicates a significant (<0.05) interaction effect between gait and sensor.

Table 1
Mean (SD) amount of points over participants per magnitude of angular change for a duration of 0.05–0.1 s of cantering. Note that this includes both flexion and extension angular changes.

	10-15°	15-20°	20-25°	>25°
PelL4 (n = 10)	163 (102)	91 (112)	34 (44)	5 (9)
L4L1 (n = 10)	59 (118)	15 (43)	0 (1)	0 (0)
L1T11 (n = 12)	2 (3)	0 (0)	0 (0)	0 (0)
PelT11 (n = 12)	124 (95)	52 (60)	16 (29)	5 (14)

walking (Ferrante et al., 2021; Kim et al., 2020) is the overall higher accelerations that are reached. Vertical acceleration peaks were generally below 5 m/s^2 (excluding g) for walking, and between 20 and 40 m/s^2 (excluding g) for cantering. We found RMS accelerations of above 1 Hz to be all higher for cantering than for walking (Fig. 4). Additionally, we found that horse riding induced large angular motions between IMUs along the lumbar spine during cantering. Pelvis-L4 angular change reached more than 20° within 0.1 s 42 (SD = 53) times on average.

4.1. RMS and transmission of linear accelerations

Two large RMS acceleration peaks were found for both gaits, with the RMS of cantering being overall significantly higher than that of walking. Peaks occurred at 1.78 and 3.56 Hz, corresponding to the gait cycle of the horse (approximately 0.6 s, 1.7 Hz, see Fig. 2), which includes several impacts due to the horse's legs hitting the ground (Funakoshi et al., 2018). Periodic movement of the body over steps is visible in the RMS at multiples of the stride frequency (Fig. 4). The overall RMS of vertical pelvis accelerations during walking (1.7 m/s^2) was similar to the results of Funakoshi et al. (2018) (1.42 m/s^2). The expected amplification of vertical accelerations at 4 Hz (Pope et al., 1996), due to the eigenfrequency of the trunk, was found in the 1.8–5 Hz frequency bands, but was relatively small. Amplification of accelerations causes an increase in proximal accelerations and therefore a higher risk of injury due to compression. For higher frequencies, vertical accelerations were damped for both gaits. These findings are in line with Kingma and van Dieën (2009). Damping of high-frequency accelerations shows that the low back functions like a low-pass filter, which is most likely related to the damping properties of the intervertebral discs (Izambert et al., 2003; Vogel and Pioletti, 2012). We found similar transmissions for walking and cantering, which may imply that the lumbar spine transmits vertical accelerations equally, regardless of the magnitude.

4.2. Compression due to linear accelerations

In vitro studies estimate the ultimate compressive strength of lumbar vertebrae at 5kN for women aged 20–40 (Jäger, 2018). With the current data, we can only estimate the compressive load as a result of gravitational and inertial forces, disregarding compression due to muscle activity and passive tissues. A roughly estimated compression load already reaches 50 % of the ultimate compressive strength (assuming upper body weight of 40 kg, 40 m/s^2 acceleration: $F = m(a + g) = 2$ kN). Any trunk inclination would increase this due to muscle forces needed to counteract the moment induced by the forces acting along the upper body centre of mass (Bazrgari et al., 2008).

It has been shown that the compression failure location depends on strain rate (Jiang et al., 2023). Additionally, the stiffness of both the intervertebral disc (Li et al., 2020; Newell et al., 2019) and the vertebral body (Kemper et al., 2013; Stemper et al., 2015) increases with strain rate. This, combined with the low RMS values and substantial damping at higher frequencies, versus the high RMS peaks and slight amplification at low frequencies, suggests that the low frequency peaks (<5 Hz) in horse riding impose the highest risk of compression failure.

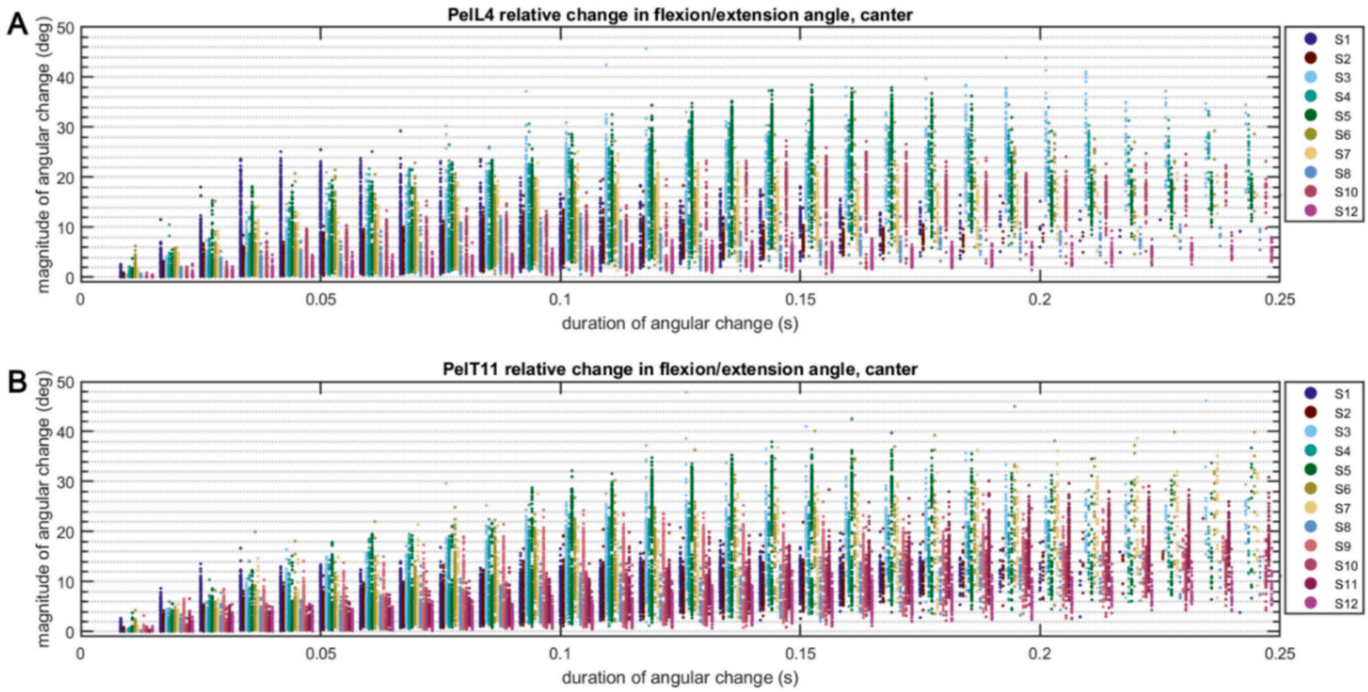

Fig. 5. Angular change between the Pel and L4 sensor (A) and between the Pel and T11 (B) sensor for all participants. The magnitude of the angular change (vertical axis) is provided per duration of the change (horizontal axis). Note that these relative angles include both extension and flexion. For visibility, the data points for each participant were offset by 0.0006 s. The true duration of each participant is therefore at the location of subject 1. The visible discretisation of the data is due to the sample frequency of 120 Hz.

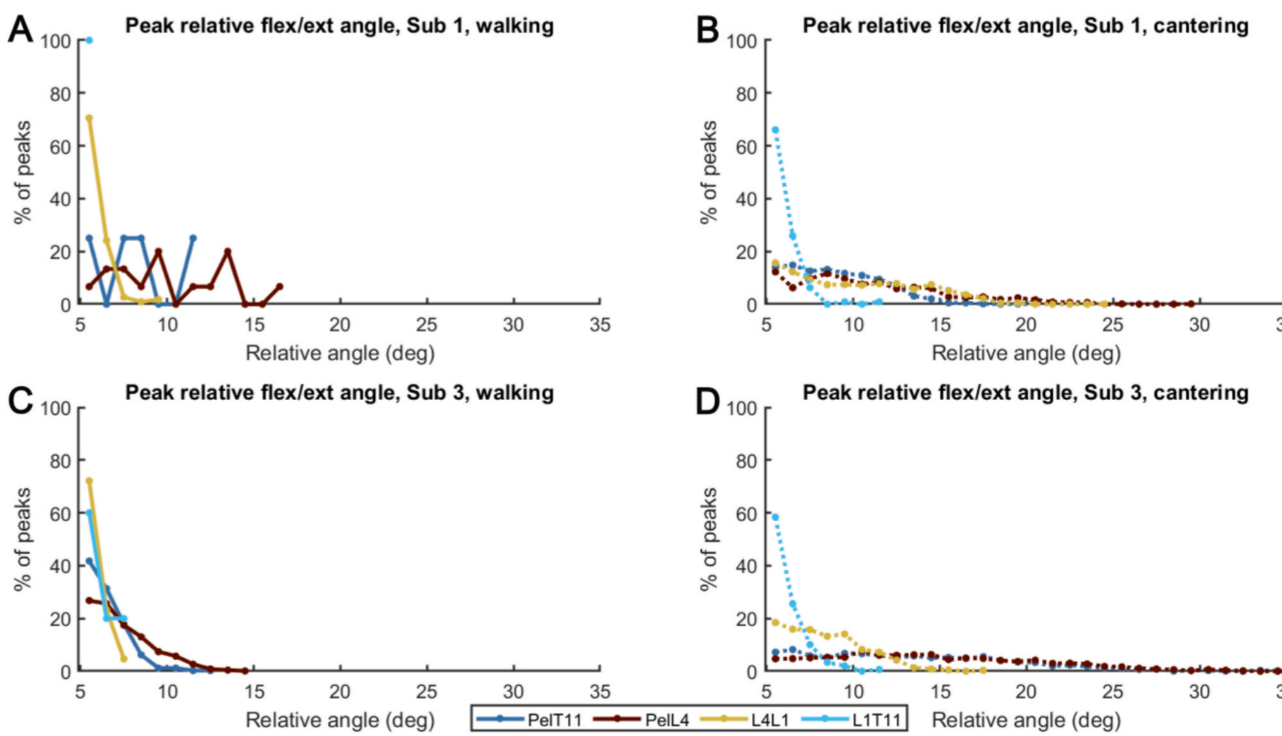

Fig. 6. Relative numbers of relative angular change peaks for walking (A, C) and cantering (B, D) for two participants. Relative number of peaks of relative angular change between measurement sites for each participant. This only includes peaks of at least 5 degrees.

4.3. Angular motions

The total lumbar (pelvis-T11) flexion–extension motion was below 30 and 40° for almost all data points in walking and cantering respectively (Figs. 5, 6 and S9). These values are well within a healthy range of motion (ROM) (flexion + extension) of 85° for adults under 30 years old (Dreischarf et al., 2014). During cantering, relative pelvis-L4 angular change was sometimes as large as 30° (Fig. 5 and S11C). These values approach the flexion ROM from upright posture, which was reported to be 26° based on *in vivo* Xray and 31° based on soft tissue injury threshold level *in vitro* (Adams and Hutton, 1986).Compared to the full flexion–extension ROM *in vivo* of 30° (Cook et al., 2015; Dreischarf et al., 2014), our numbers are still somewhat lower. However, it should be noted that IMU movement differs from vertebral movement because of possible IMU movements relative to the vertebrae. Those large movements between the pelvis and L4 sensor were mainly induced by pelvic rotation. Pelvic rotation in cantering was on average 26.3° in our study, which is somewhat larger than previously reported values of 18° during cantering (Münz et al., 2014) and 9.7° (Münz et al., 2014) to 11.1° (Byström et al., 2010) during dressage riding.

While we were unable to distinguish between voluntary and involuntary lumbar motion or buckling, we consider the relatively large angular movements of the equestrian at the stride frequency (1.7 Hz, Fig. 2B) to be likely voluntary, in order to align their motion with the motion of the horse. Movements at higher frequencies might be involuntary as an effect of accelerations due to the impact of the horse's hoofs striking the ground and the buttocks colliding with the saddle. For cantering, we found large angular changes (>20°) within very short time intervals (<0.1 s), which were not always obvious from the angular acceleration (e.g., see Figs. 2B, 2F, t = 200.9–201.00). Additionally, we found large, brief angular acceleration peaks sometimes resulting in only minor angular changes (e.g., see Figs. 2B, 2F, t = 200.8).

Cholewicki et al. (1999) showed in a model study that the critical load causing buckling of the spine depends on muscular co-activation and on intra-abdominal pressure. Around the neutral posture, there is a neutral zone where spine stiffness is low (Wilke et al., 2025) and therefore muscular co-contraction is needed to avoid buckling, even without external impact loads (Cholewicki et al., 1997). In terms of injury risk, on the one hand, more rapid angular motions are known to stiffen the lumbar spine (Stolworthy et al., 2014; Wagnac et al., 2012). On the other hand, substantial angular motions within 0.1 s. are too fast for the neuromuscular system to allow for voluntary reactive muscle activation, or even for reflexes, to counteract moments inducing these angular motions. For instance, sudden release experiments showed an antagonist EMG response time of 70 ms (Radebold et al., 2000), and the electromechanical delay adds about 94 ms to that in sitting posture. Therefore, anticipatory muscular co-activation would be needed to avoid injury if stiffness is not sufficient to avoid overstraining tissues.

For some participants, fast angular changes appeared to be present while the spine was already in a bend position (Figs. S11-S12). While the spine is much stiffer close to its end range than around its neutral zone (Cholewicki and Mcgill, 1996; S. McGill et al., 1994), small additional rotations could induce overstraining of tissues. In this context, it should also be noted that, close to the end range of flexion, application of vertical impact forces shows reduced failure strength (Campbell-Kyureghyan et al., 2011) and a modified failure mode (Cutlan et al., 2024). The ROM of spine segments is limited by posterior and anterior ligaments, as well as by the facet joints (Heuer et al., 2007).

4.4. Limitations

We measured movements of skin-mounted IMUs, rather than those of vertebrae. Movement of IMUs and markers relative to the skin, and soft tissue deformation (e.g. muscle contraction) and movement (e.g. joint angular change) between skin and bone have been the subject of extensive discussions (Croce, 2006; Cutti et al., 2005; Leardini et al., 2005; Rong, 2022). To obtain a crude estimate of these effects, we conducted two brief measurements (see Appendix): (1) IMUs were attached to either side of a rigid rod and tapped on a surface, revealing relative angular movements of 1–2° (Fig. A2); and (2) IMUs were attached to either side of a forearm and the elbow was tapped, revealing relative angular changes of 1.5–2° (Fig. A3). When comparing the skin to tissue motion of the forearm to the low back, the errors on the low back are likely comparable to or smaller than 2°.

However, skin movement over bone may still be relevant for measurements of relative angles in places where IMUs span multiple joints. For example, an IMU placed on T11 in an upright position may shift caudally during flexion. Additionally, IMUs are placed using palpation and estimation, and the exact location of the IMUs relative to the vertebrae is unknown. Moreover, orientation changes of the skin of the low back during lumbar motion may not fully reflect angular changes of vertebrae. Clearly, further validation is needed, preferably with fluoroscopy (Teyhen et al., 2005; Wong et al., 2004).

Because of the relatively small sample size and high inter-participant variability, statistical power of our study is limited. Moreover, L4 sensor data of two participants was missing due to technical difficulties, leading to a smaller sample size for L4-dependent variables. However, the relative angular changes among the remaining sensors were consistent with those of other participants. Therefore, we do not expect these missing data to skew the outcome.

Finally, only female equestrians volunteered to participate in this study. This is likely the result of a relatively small number of male equestrians in the Netherlands. While demographic variables such as sex and age have been shown to affect lumbar spine stiffness (Sawa et al., 2020), it is unclear if this may have affected the results. Future work should investigate the lumbar kinematics and transmission of acceleration in male participants, in experienced vs inexperienced equestrians, and potential effects of adaptation in those who have, or have had, low back pain. For instance, it is conceivable that inexperienced equestrians would have less smooth adaptation to the horse's movement, resulting in higher peak acceleration due to larger impact forces of the buttocks against the saddle. Also, the voluntary motion and timing and magnitude of bracing of trunk muscles could depend on experience or LBP history, potentially affecting effects of acceleration peaks on kinematics. Furthermore, the vertebral compressive strength is partially dependent on bone mineral density, which is influenced by loading. Hence, trained equestrians are likely more resistant to vertebral fractures due to impact than untrained equestrians.

5. Conclusion

We found vertical accelerations up to about 5G (including gravity) during cantering, with a slight amplification below 5 Hz and damping at higher frequencies. Furthermore, we found large and fast local angular motions, with an average of 43 (SD = 53) times over 20° within 0.1 s between the Pelvis and L4 sensors. Despite an unknown dose-risk relationship, these large and fast local angular changes during horse riding may pose a risk for ligament damage, especially when they take place near the end of the ROM. It is therefore essential to consider angular changes when assessing the effect of repeated shocks on the low back.

In conclusion, evaluating the effect of repeated shocks in terms of linear accelerations alone seems to be insufficient. Understanding the risk of injury due to buckling is crucial for accurate evaluation of the effect of repeated shocks on the low back. In that respect the effects reported in this paper require further experimental and/or model-based validation.

CRediT authorship contribution statement

N.A. Smit: Writing – review & editing, Writing – original draft, Visualization, Validation, Supervision, Software, Resources, Project administration, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **J.E. Bos:** Writing – review & editing, Supervision, Methodology, Funding acquisition, Formal analysis, Data curation, Conceptualization. **J.H. van Dieën:** Writing – review & editing, Formal analysis, Conceptualization. **I. Kingma:** Writing – review & editing, Supervision, Software, Resources, Project administration, Methodology, Formal analysis, Data curation, Conceptualization.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2025.113067.

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