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# First steps towards reducing chronic low back pain in horseback riders: Objectifying biomechanical parameters using inertial sensors

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# SAMENVATTING

Paardrijden is wereldwijd een populaire sport voor ruiters van alle leeftijdscategorieën. Helaas is de prevalentie van chronische lage rugpijn (CLBP) bij ruiters significant hoger dan in de algemene populatie. Ondanks verschillende onderzoeken is nog niet duidelijk waar deze hoge prevalentie vandaan zou kunnen komen. Discipline, trainingsintensiteit en ervaring lijken geen duidelijke invloed te hebben. Uit een vergelijkend onderzoek met Magnetic Resonance Imaging (MRI) kwamen ook geen veranderingen in het spierkorset of in de tussenwervelschijven naar voren. Bij andere sporten lijkt er een relatie te bestaan tussen piekversnellingen en schokdemping en CLBP.

De bewegingen van een persoon worden vaak geanalyseerd met behulp van optische meettechnieken. In de laatste jaren worden er naast deze optische technieken ook vaak inertiële meetsensoren (IMUs) gebruikt. IMUs hebben als voordeel dat ze goedkoper zijn en vrijwel overal gebruikt kunnen worden. Het doel van deze studie was om IMUs te gebruiken om biomechanische parameters die mogelijk een relatie hebben met CLBP (piekversnellingen en schokdemping) te meten bij ruiters te paard. Bij ruiters is al eerder gemeten met IMUs, maar deze specifieke parameters zijn nog niet vaak in kaart gebracht. Daarnaast gebruiken de meeste studies waarin ruiters gemeten worden met IMUs een extra sensor op het paard om de gangcyclus te detecteren. Een tweede doel van deze studie was dan ook om de gangcyclus van het paard te detecteren op basis van de metingen van de ruiter.

Tien gezonde vrouwelijke ruiters (leeftijd: 27.3 ± 5.8 jaar [gemiddelde ± SD]; lengte: 173.7 ± 6.5 cm, gewicht: 73.4 ± 13.8 kg, BMI: 24.3 ± 4.2, rijervaring: 18.7 ± 6.0 jaar) hebben deelgenomen aan dit onderzoek. De metingen zijn uitgevoerd met het MVN Link systeem, waarbij het Lycra pak werd gebruikt om acht sensoren te bevestigen (op de voeten, onderbenen, bovenbenen, pelvis en sternum). Alle deelnemers legden op hun gebruikelijke paard op de linkerhand (tegen de klok in) het volgende protocol af: 3 ronden stap, 3 ronden draf doorzitten, 3 ronden galop, 3 ronden stap zonder beugels, 3 ronden draf doorzitten zonder beugels, 3 ronden galop zonder beugels.

Detectie van de gangcyclus van het paard op basis van de heuphoeken van de ruiter was succesvol in draf doorzitten en galop. In stap waren de bewegingen veel kleiner en was er bij vijf ruiters geen duidelijk patroon in de heuphoeken van de ruiters te detecteren. Voor drie ruiters kon dit opgelost worden door de hoeken van het lumbosacrale gewricht te gebruiken. Voor twee ruiters bleef het moeilijk om een duidelijk patroon te detecteren en moest dit door middel van visuele inspectie van de data met de hand gebeuren. De absolute waarden van de versnellingen in stap waren laag (onder de 2 $m/s^2$). Hierdoor was het moeilijk om een patroon in de versnellingswaarden van stap te ontdekken en is deze data niet verder geanalyseerd.

In draf doorzitten en galop waren wel duidelijke versnellingspatronen zichtbaar. In galop is een patroon met één hoge piek zichtbaar. In draf doorzitten is een patroon met twee pieken van gelijke hoogte zichtbaar per gangcyclus, waarbij opviel dat de piekversnelling in het sternum significant hoger was dan de piekversnelling in het pelvis. Hier werd van pelvis naar sternum dus een negatieve schokdemping gevonden. Dit is tegen de verwachtingen, aangezien schokdemping-strategieën gericht zijn op het zo klein mogelijk houden van de schokken hoger in het lichaam. Een casusstudie met een ander meetsysteem (Xsens DOT) suggereert dat er een roterende component aanwezig is in het pelvis die niet geheel doorgegeven wordt aan het sternum. Dit zou de hogere versnelling in het sternum kunnen verklaren en ook van belang kunnen zijn bij de ontwikkeling van CLBP. Voor de schokdemping van de voet naar het onderbeen werden in de condities met beugels positieve waarden gevonden, wat betekent dat de impact gedempt wordt. In de condities zonder beugels werd de schokdemping significant lager of zelfs negatief, wat suggereert dat het gebied tussen de voet en het onderbeen in de conditie met beugels belangrijk is om de schok te dempen. Het rijden zonder beugels leidde, ondanks de verminderde schokdemping in het onderbeen, niet tot hogere versnellingen in de torso.

De resultaten van deze studie suggereren dat de rompstabiliteit en rompflexibiliteit van ruiters van belang zouden kunnen zijn met betrekking tot het al dan niet ontwikkelen van CLBP. De huidige studie is gedaan met gezonde, jonge vrouwelijke ruiters. Mogelijk hebben oudere ruiters en ruiters met overgewicht een grotere kans op het ontwikkelen van CLBP vanwege, respectievelijk, verminderde rompflexibiliteit en rompstabiliteit.

# ABSTRACT

Horseback riding is a popular sport worldwide amongst all age categories. However, the prevalence of chronic low back pain (CLBP) is significantly higher amongst horseback riders compared to the general population. Despite the efforts of several research groups looking into the effects of riding discipline, riding intensity and riding experience, there is no clear explanation for this high prevalence yet. A study using Magnetic Resonance Imaging (MRI) found no indications of muscular changes or damage to the intervertebral discs in riders experiencing CLBP. Research in other sport disciplines suggests there might be a relationship between CLBP and peak accelerations and shock attenuation.

Optical motion capture systems are typically used to analyze the movements of an individual. In recent years, however, inertial measurement units (IMUs) are more frequently being used to objectify the movements of an individual. IMUs have the advantage of being more cost-friendly and, unlike motion capture systems, can be used in practically any environment. The goal of the current study was to use IMUs to measure biomechanical parameters that are likely to be related to CLBP (peak acceleration and shock attenuation) in horseback riders. IMUs have been used to measure horseback riders before, but, to the best of our knowledge, these specific parameters have not yet been clearly objectified before. Furthermore, most studies in which IMUs are used in horseback riding use an additional sensor located on the horse to detect the gait cycle of the horse. A secondary goal of this study was to detect the gait cycle of the horse based solely on the data derived from the rider.

Ten healthy female horseback riders (age: 27.3 ± 5.8 years [mean ± SD]; height: 173.7 ± 6.5 cm, weight: 73.4 ± 13.8 kg, BMI: 24.3 ± 4.2, riding experience: 18.7 ± 6.0 years) participated in this study. The measurements were performed with the MVN Link system. The Lycra suit was used to attach eight sensors (located at the feet, lower legs, upper legs, pelvis and sternum) to the participant. All participants rode their usual horse and completed the following protocol on the left lead (counterclockwise): 3 rounds of walk, 3 rounds of sitting trot, 3 rounds of canter, 3 rounds of walk without stirrups, 3 rounds of sitting trot without stirrups, 3 rounds of canter without stirrups.

Gait cycle detection based on hip angles of the rider was successful for sitting trot and canter. In walk, the movements of the horse are smaller and gait cycle detection was difficult for five riders. For three riders, gait cycle detection in walk was possible by using the lumbosacral joint angles instead of the hip angles. For two riders, gait cycle detection remained difficult in walk and was performed manually by means of visual inspection of the data. For all riders, the absolute acceleration values in walk were low (below 2 $m/s^2$), making it difficult to detect a clear pattern in the acceleration values. Therefore, the walk data was not analyzed further.

A clear acceleration pattern was visible for sitting trot and canter. In canter, an acceleration pattern with one large peak was visible. In sitting trot, two peaks of equal height were visible per gait cycle. In sitting trot, a negative shock attenuation between the pelvis and sternum was found, with the acceleration in the sternum being significantly higher than the acceleration in the pelvis. This is contrary to expectations, as shock attenuation is normally aimed at keeping superior impacts minimal. A case study with a different measurement system (Xsens DOT) suggests that there might be a rotational component in the pelvis that is not fully transferred to the sternum. This could explain the higher acceleration values found in the sternum and could also be of interest with regards to the development of CLBP. Shock attenuation from the foot to the lower leg was positive for all conditions with stirrups, meaning that the shock was attenuated. In conditions without stirrups, shock attenuation was significantly lower or even negative. This suggests that the area between the foot and the lower leg is important in shock attenuation when riding with stirrups. Riding without stirrups did not lead to higher accelerations in the torso, despite the decreased shock attenuation in the lower leg.

The results of this study suggest that the flexibility of the torso and core stability could be of importance with regards to the development of CLBP in horseback riders. The current study was performed with young, healthy female riders. Older or obese riders might have an increased risk of developing CLBP due to a diminished flexibility of the torso and a diminished core stability, respectively.

## SYMBOLS AND ACRONYMS

| Symbol / Acronym | Definition | Unit |
|---|---|---|
| $\boldsymbol{a}$ | Acceleration | $m/s^2$ |
| $\boldsymbol{a}_i$ | Acceleration along the specified axis | $m/s^2$ |
| $\lvert\boldsymbol{a}\rvert$ | Modulus of the acceleration | $m/s^2$ |
| **CLBP** | Chronic low back pain | |
| **IMU** | Inertial measurement unit | |
| **MRI** | Magnetic Resonance Imaging | |
| **P - S** | Pelvis – Sternum (shock attenuation) | |
| **PEL** | Pelvis | |
| **RFO – RLL** | Right foot – Right lower leg (shock attenuation) | |
| **RFO** | Right foot | |
| **RLL** | Right lower leg | |
| **RUL** | Right upper leg | |
| **S** | Stirrups | |
| **STE** | Sternum | |
| **WS** | Without stirrups | |

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# 1. INTRODUCTION

With 37 million individuals riding horses at least once a week (1), horseback riding is a widely practiced and popular sport worldwide amongst all age categories. Even though it is the only sport in which men and women can compete directly throughout all levels, horseback riders are predominantly female (1). Besides the beneficial effects of horseback riding on for example balance and motor control, a common complaint in horseback riders is chronic low back pain (CLBP). It has been shown that the prevalence of CLBP is significantly higher in horseback riders (88%) compared to controls (33%) (2). Other studies found a lifetime prevalence of back pain of 81% (3) and an incidence of 72.5% (4) in horseback riders, with the pain mainly being located in the lumbar spine. To minimize injury risk and associated costs (5), risk factors and potential causes for CLBP in horseback riders should be identified and understood.

The movements of the horse create a mechanical impact that is transferred from the horse to the rider. Shock attenuation in the human body is aimed at keeping the accelerations for the proximal part of the human body minimal. This is achieved through both passive and active mechanisms (6). In a simplified model in which only the lower extremities and trunk are considered, there are three main points of impact for a horseback rider: one at each foot, where the shock is transferred via the stirrups, and one at the pelvis, where the shock is transferred via the saddle. The impacts at the feet can be attenuated throughout the lower leg before it reaches the pelvis and the trunk. However, the impact transferred via the saddle has a more direct impact on the lower back. Part of this impact is likely to be absorbed by the intervertebral discs in the spinal cord (7). Damage to the intervertebral discs could thus potentially explain CLBP in horseback riders. However, in a study wherein horseback riders (40 females, 18 males; mean age: 32.4 years) experiencing CLBP were evaluated by means of Magnetic Resonance Imaging (MRI), no changes in paraspinal musculature or degeneration of the intervertebral discs were found (2). Other studies looked at the effect of riding discipline, riding intensity and experience (3,4). Whereas a smaller study including 32 horseback riders (22 females, 10 males, mean age: 25 years) found a higher prevalence of CLBP in professional horseback riders and in jumping riders (4), a larger scale study including 508 horseback riders (321 females, 187 males, mean age: 33.5 years) found no significant impact of riding intensity and discipline of the frequency or severity of back pain (3). The results of these studies in the potential causes for CLBP are somewhat conflicting. Thus, potential causes for CLBP in horseback riders have yet to be clearly identified.

There seems to be more consensus with regards to potential factors involved in CLBP in other cyclical movements and sports. Joint stiffness and shock attenuation properties have been linked to CLBP in walking and running (8,9). Joint stiffness in the lower extremities, especially the knee joints, seems to be connected to CLBP in runners and baseball players (9,10). It is hypothesized that the increased joint stiffness alters the shock attenuation properties of the lower leg, thereby exposing the lower back to larger mechanical shocks. Although the causality of these factors has not been investigated in these studies, there does seem to be a correlation between CLBP and shock attenuation in the lower extremities and pelvis. To the best of our knowledge, shock attenuation has not yet been measured in horseback riders. A similar relationship between shock attenuation and CLBP might exist in horseback riding.

Traditionally, the biomechanics of horseback riders have been studied using optical motion capture systems (11-13). The use of this technique is limited by the need for an elaborate and expensive camera system that needs to be placed in the riding arena. As an alternative, inertial measurement units (IMUs) can be used to measure the rider (14-21). An IMU uses accelerometers, gyroscopes and sometimes magnetometers to measure the acceleration and angular velocity of the body or object to which it is attached. By placing these sensors on the rider at well-chosen location, the acceleration and angular velocity of the body parts of the rider can be measured. These parameters can then be used to estimate other (biomechanical) parameters, such as the position and the orientation. IMUs can be used in practically any environment, thereby improving the applicability compared to optical motion capture systems. In most studies performed with IMUs on horseback riders, an extra sensor is placed somewhere on the horse to define the gait cycle. One study, for example, used an IMU on the sternum of the horse and an IMU on the pelvis of the rider to measure the kinematics of both horse and rider (15). The IMU located on the horse was used to define the gait cycle of the horse, and the movements of the rider and horse were compared (15). Other topics studied with horseback riders and inertial sensors include horse-rider symmetry, hip rotation asymmetry, pelvis kinematics and the effects of stirrups length (15,17-19,21). Although these studies often investigate kinematic parameters, they rarely discuss the biomechanical implications of the found values for the rider in relation to injuries. Furthermore, most studies use two measurement systems, one for the horse and one for the rider, which complicates the set-up by requiring synchronization of the measurement systems.

Since peak accelerations and shock attenuation have been linked to CLBP in other sports, the primary goal of this study was to use IMUs to objectify these biomechanical parameters in healthy horseback riders. To the best of our knowledge, IMUs have not yet been used to objectify shock attenuation in horseback riders. It was hypothesized that peak accelerations and shock attenuation can be measured using IMUs, based on previous studies performed in runners (6) and horseback riders (21). It was also hypothesized that peak accelerations will decrease in magnitude in segments superior to the impact point. The secondary goal of this study was to detect the gait cycle of the horse solely based on the data derived from inertial measurements of the rider. It was hypothesized that pelvic angles of the rider could be used to detect the gait cycle of the horse (15).

# 2. METHODS

## 2.1 Participants

Ten female horseback riders between 21 and 40 years of age (age: 27.3 ± 5.8 years [mean ± SD]; height: 173.7 ± 6.5 cm, weight: 73.4 ± 13.8 kg, BMI: 24.3 ± 4.2, riding experience: 18.7 ± 6.0 years) with a minimum level comparable to the Dutch L-level in dressage participated in the study. Participants were excluded if they (a) normally rode less than 3 hours per week, (b) had major injuries in the last 6 months, (c) had surgery in the lower extremities and trunk, (d) experienced start- or morning-stiffness, (e) had (lower) back pain or (f) were pregnant. The participants rode on their regular horses (6 Dutch warmbloods (KWPN) and 1 German warmblood (Holsteiner), age: 14.7 ± 5.3 years, height at withers: 169.1 ± 3.3 cm). Horses were excluded if they showed any signs of lameness. All participants signed an informed consent before participation.

## 2.2 Experimental design

### 2.2.1 Motion capture

Participants wore a Lycra suit in which 8 IMUs (MTx, Xsens Technologies BV, Enschede, The Netherlands) were placed on pre-determined position (see Figure 1 and Figure 2). This Lycra suit is part of a full-body motion capture system (MVN Link) and was used in the ‘lower extremities and trunk’ set-up, with IMUs being placed at the feet, lower legs, upper legs, pelvis and sternum. The sternum sensor was not placed in the Lycra suit but taped to the skin. This was done to prevent movement between the sensor and the segment, which was found to be more present in the sternum sensor during a pilot study with the Lycra suit. The IMUs consisted of 3D accelerometers, 3D gyroscopes and 3D magnetometers, all sampling at 240 Hz. Anthropometric data was collected from the participants to ensure a representative biomechanical model was created by the MVN Analyze software (Xsens Technologies BV, Enschede, The Netherlands). The body dimensions measured were body height, shoe length, shoe height, hip height, hip width, knee height and ankle height. Before the measurement, participants


*Figure 1: Sensor set-up of the MVN Link system with riding breeches and boots worn over the Lycra suit.*

performed a calibration procedure in which they were instructed to stand in a neutral pose and walk back and forth (approximately 10 meters in total) in a straight line. Since the data was collected offline, the calibration procedure was performed three times to ensure a good quality calibration was present for each participant. Regular horse-riding clothes and boots were worn over the Lycra suit and motion capture equipment to mimic the normal riding situation as closely as possible.


*Figure 2: Sensor set-up of the MVN Link system with riding breeches and boots worn over the suit, as the participant is sitting on the horse.*

### 2.2.2 Procedure

Data collection took place in an indoor riding arena with a minimum size of 20 by 40 meters. Participants were instructed to warm-up their horse as they would normally do. After participants had performed their self-selected warm-up in which they had at least walked, trotted and cantered, the riders were asked to start with the study protocol. They were instructed to start on the short side of the riding arena and stand still for a few seconds. The participants then performed three laps of walk, three laps of sitting trot, three laps of canter, three laps of walk without stirrups, three laps of sitting trot without stirrups and three laps of canter without stirrups on the left rein (counterclockwise). For every transition between paces, the participants were instructed to stand still at the short side (their starting point) for a few seconds. Participants were instructed to keep the pace as steady as possible for every condition. A graphic overview of the measurement protocol can be found in Figure 3. A videorecording was made during the execution of the protocol so that potential unexpected or sudden movements from the horse could be excluded in the analysis phase.

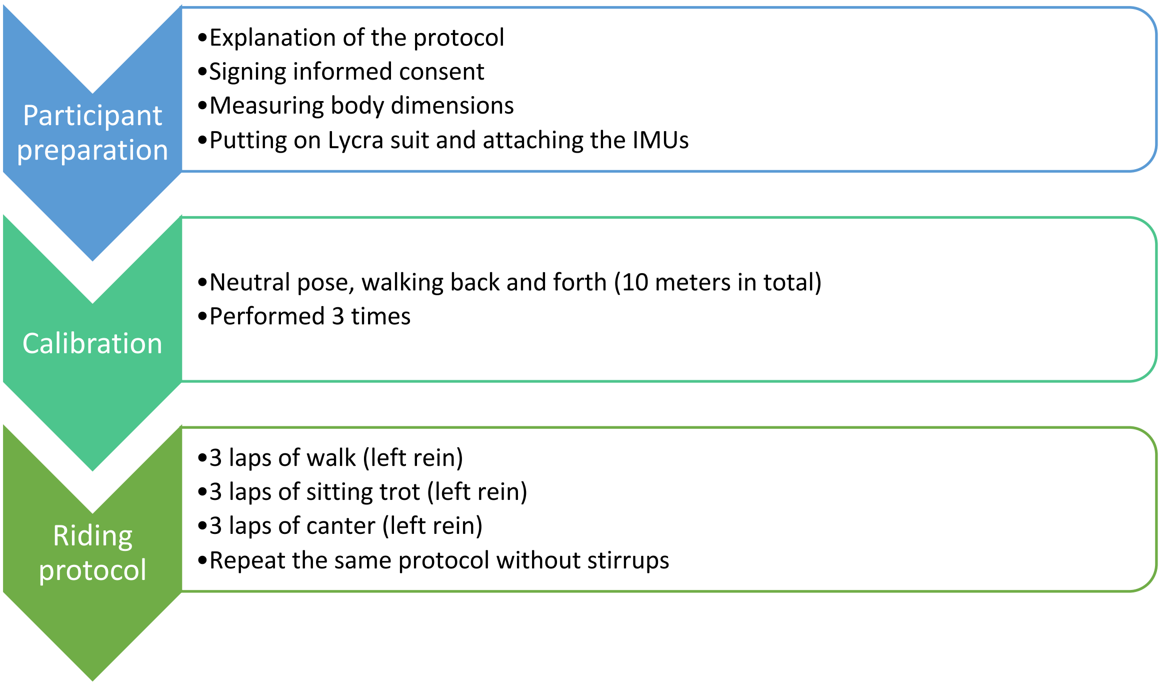


*Figure 3: Graphic overview of the measurement protocol.*

## 2.3 Analysis

The anthropometric data from the participants was entered into the MVN Analyze software (version 2019.2). Data was processed on a single level and exported for further analysis in MATLAB 2020a (MathWorks Inc., Natick, MA). Since participants were riding on the left rein, the left leg was more involved in giving aids to the horse. These aids are given by means of active muscle contractions and subsequent movements of the leg of the rider, which could influence the measurements of the IMUs. Therefore, only the measurements of the right lower extremity were used for data analysis. The segments that were used for analysis were the right foot, right lower leg, right upper leg, pelvis and sternum as defined by MVN (22). Data was analyzed in the global frame, in which the x-axis point to the local magnetic North, the z-axis points up and the y-axis is defined according to a right-handed coordinate system.

Using MATLAB 2020a, 20 gait cycles were selected and analyzed per participant per condition (walk, sitting trot and canter with and without stirrups). The gait cycles of the horse were selected based on hip and pelvic (defined as the movement in the lumbosacral joint) angles of the rider (15). Due to the nature of the movement, two peaks per gait cycle occur in walk and trot, and one peak per gait cycle occurs in canter (15). Figure 4 shows the movement patterns of the horse. Walk is a 4-beat movement, where there are two moments per gait cycle in which the hindleg creates a forward motion, translating to two peaks in the pelvic or hip angles of the rider (15). In trot, the 2-beat rhythm of the horse leads to a movement pattern in the rider where two peaks in the pelvic or hip angles are visible per gait cycle (15). Canter is a 3-beat movement, where the rolling movement of the horse leads to one peak in the pelvic or hip angles of the rider (15).

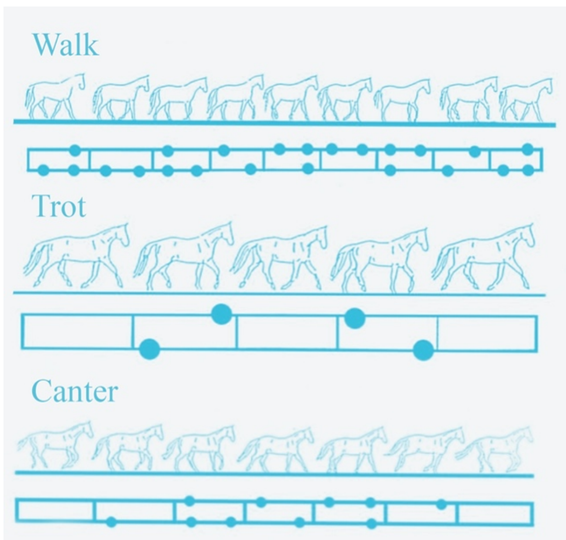


*Figure 4: Movement patterns of the horse in walk, trot and left lead canter (23).*

The acceleration data was analyzed in two ways. Firstly, the data was analyzed by taking only the vertical acceleration into account. We hypothesized that this could give an impression of the compression forces present in the lower extremities and torso. Secondly, the data was analyzed by looking at the modulus, thereby also encompassing acceleration in other directions. The modulus of the acceleration (in m/s$^2$) is defined as:

$$|a| = \sqrt{a_x{}^2 + a_y{}^2 + a_z{}^2} \quad (m/s^2) \tag{1}$$

Where $|a|$ denotes the modulus of the acceleration, and $a_x$, $a_y$ and $a_z$ are used to denote the accelerations along the x-, y-, and z-axes respectively.

The data was time normalized using linear interpolation. Peak accelerations were determined and used to calculate shock attenuation (in %), which was defined as:

$$Shock\ attenuation\ (peak) = \left(1 - \frac{peak\ acceleration\ superior\ segment}{peak\ acceleration\ inferior\ segment}\right) \cdot 100\ \ (\%) \tag{2}$$

If more peaks of equal height were present per gait cycle, as is expected for sitting trot, the average of these peaks was used as the peak acceleration of the segment. Shock attenuation was calculated from the right foot to the right lower leg and from the pelvis to the sternum. The right upper leg is already in contact with the saddle, which would have limited the informativeness of a shock attenuation parameter involving the right upper leg.

Using the gait cycles in the time domain, the area under the acceleration curve was calculated to represent the total impact or shock. Shock attenuation (in %) was calculated using the areas under the curve as:

$$Shock\ attenuation\ (area) = \left(1 - \frac{area\ superior\ segment}{area\ inferior\ segment}\right) \cdot 100\ \ (\%) \tag{3}$$

Shock attenuation was again calculated from the right foot to the right lower leg and from the pelvis to the sternum. Statistical analysis was performed using IBM SPSS Software (version 27.0.1.0) (IBM Analytics, Armonk, NY). Normality of the data was checked by means of the Shapiro-Wilk test and inspection of the Q-Q plots. Repeated-measures ANOVA was used to test whether any significant differences were present in the peak accelerations for the conditions with and without stirrups.

## 3. RESULTS

Gait cycle detection based on hip angles was successful for all participants in sitting trot and canter with and without stirrups. For 3 participants, pelvic angles instead of hip angles were used to detect the gait cycle in walk. For 2 participants, gait cycle detection was difficult in walk. For all participants, the absolute acceleration values in both conditions in walk were low (below 2 m/s$^2$), with a large variation both inter- and intrapersonal. No clear acceleration pattern was visible for the accelerations in the vertical direction and for the modulus of the acceleration (see Appendix A). Therefore, shock attenuation was not calculated for walk.

In the vertical direction in sitting trot with stirrups, a clear acceleration pattern with two peaks per gait cycle was visible (see Figure 5). The acceleration in the sternum is significantly higher than the acceleration in the pelvis. This holds true for both the peak value ($p = 0.007$, paired samples t-test) as the area under the curve ($p=0.001$, paired samples t-test). The peak in the sternum seems to be slightly later in the gait cycle compared to the peak in the pelvis. Accelerations patterns do not seem to differ when riding without stirrups (see Appendix B). The acceleration values in the right upper leg are larger than expected with a less smooth pattern. A similar acceleration pattern is visible for the modulus of the acceleration, with the negative peaks being reversed due to the mathematical implication of taking the modulus (see Appendix B). Shock attenuation for the conditions with and without stirrups can be found in Table 1 and Table 2 for the peak and area calculations respectively.

In canter, an acceleration pattern with one large and one small peak is visible. Figure 6 shows the accelerations in the vertical direction. Similar acceleration patterns are visible for the condition without stirrups (see Appendix C). When taking the modulus of the acceleration, the smaller peak in the acceleration pattern becomes less visible (see Appendix C). Similar to sitting trot, the acceleration values in the right upper leg are larger than expected with a less smooth pattern, especially for the modulus. Shock attenuation was calculated based on the large peak. Shock attenuation for the conditions with and without stirrups can be found in Table 1 and Table 2 for the peak and area calculations respectively.

*Table 1: Shock attenuation values (%) for sitting trot and canter calculated based on the peak accelerations, represented as mean ± standard deviation. RFO: right foot, RLL: right lower leg, RUL: right upper leg, PEL: pelvis, STE: sternum, S: conditions in which the participants were riding with stirrups, WS: conditions in which the participants were riding without stirrups. A significance level of 0.05 was used, an asterisk is used to denote significant differences.*

| | | **Peak acceleration: shock attenuation vertical direction (%)** | | | **Peak acceleration: shock attenuation modulus (%)** | | |
|---|---|---|---|---|---|---|---|
| | | S | WS | p-value | S | WS | p-value |
| **Sitting trot** | RFO - RLL | 24.9 ± 8.6 | 14.9 ± 7.1 | **0.011*** | 27.7 ± 12.0 | 23.2 ± 7.9 | 0.332 |
| | P - S | -26.0 ± 29.9 | -20.2 ± 36.7 | 0.701 | -46.4 ± 66.8 | -45.3 ± 85.2 | 0.975 |
| **Canter** | RFO - RLL | 7.9 ± 18.2 | -14.3 ± 26.6 | **0.043*** | 11.8 ± 8.9 | 6.6 ± 12.8 | 0.299 |
| | P - S | -2.7 ± 22.4 | -0.9 ± 32.8 | 0.891 | -2.5 ± 46.3 | -3.4 ± 70.3 | 0.974 |

*Table 2: Shock attenuation values (%) for sitting trot and canter calculated based on the area under the curve, represented as mean ± standard deviation. RFO: right foot, RLL: right lower leg, RUL: right upper leg, PEL: pelvis, STE: sternum, S: conditions in which the participants were riding with stirrups, WS: conditions in which the participants were riding without stirrups. A significance level of 0.05 was used, an asterisk is used to denote significant differences.*

| | | **Area: shock attenuation vertical direction (%)** | | | **Area: shock attenuation modulus (%)** | | |
|---|---|---|---|---|---|---|---|
| | | S | WS | p-value | S | WS | p-value |
| **Sitting trot** | RFO - RLL | 26.9 ± 7.5 | 22.7 ± 5.6 | 0.170 | 24.6 ± 8.0 | 20.1 ± 7.6 | 0.212 |
| | P - S | -1.8 ± 8.6 | -0.4 ± 12.3 | 0.776 | -4.6 ± 25.9 | -7.2 ± 30.3 | 0.839 |
| **Canter** | RFO - RLL | 8.9 ± 7.4 | 0.3 ± 6.1 | **0.011*** | 11.6 ± 10.9 | 8.2 ± 12.0 | 0.513 |
| | P - S | 8.2 ± 6.5 | 10.4 ± 7.1 | 0.488 | 12.5 ± 15.0 | 12.9 ± 18.0 | 0.963 |

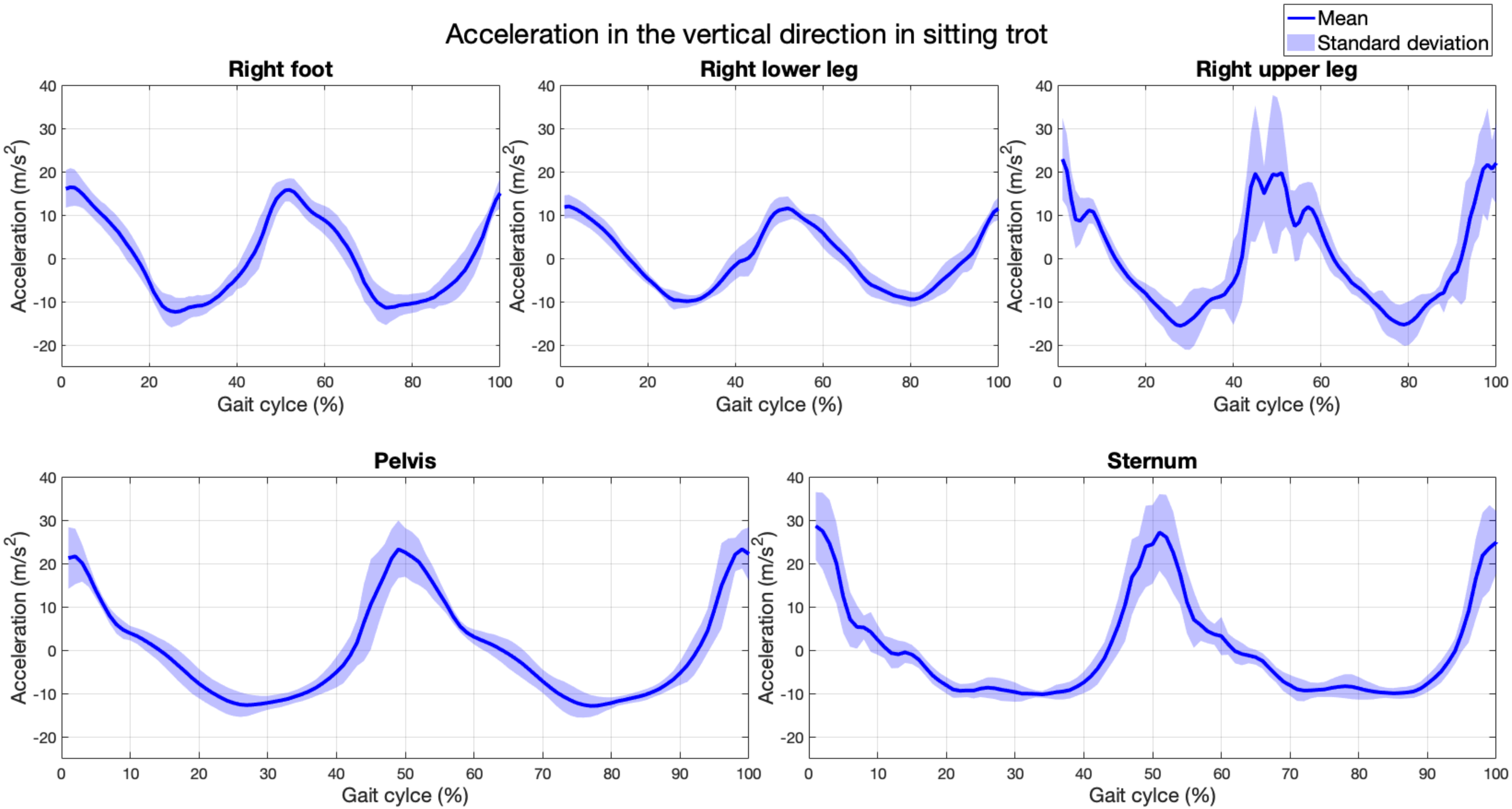


*Figure 5: Acceleration in the vertical direction in sitting trot with stirrups for the different segments.*

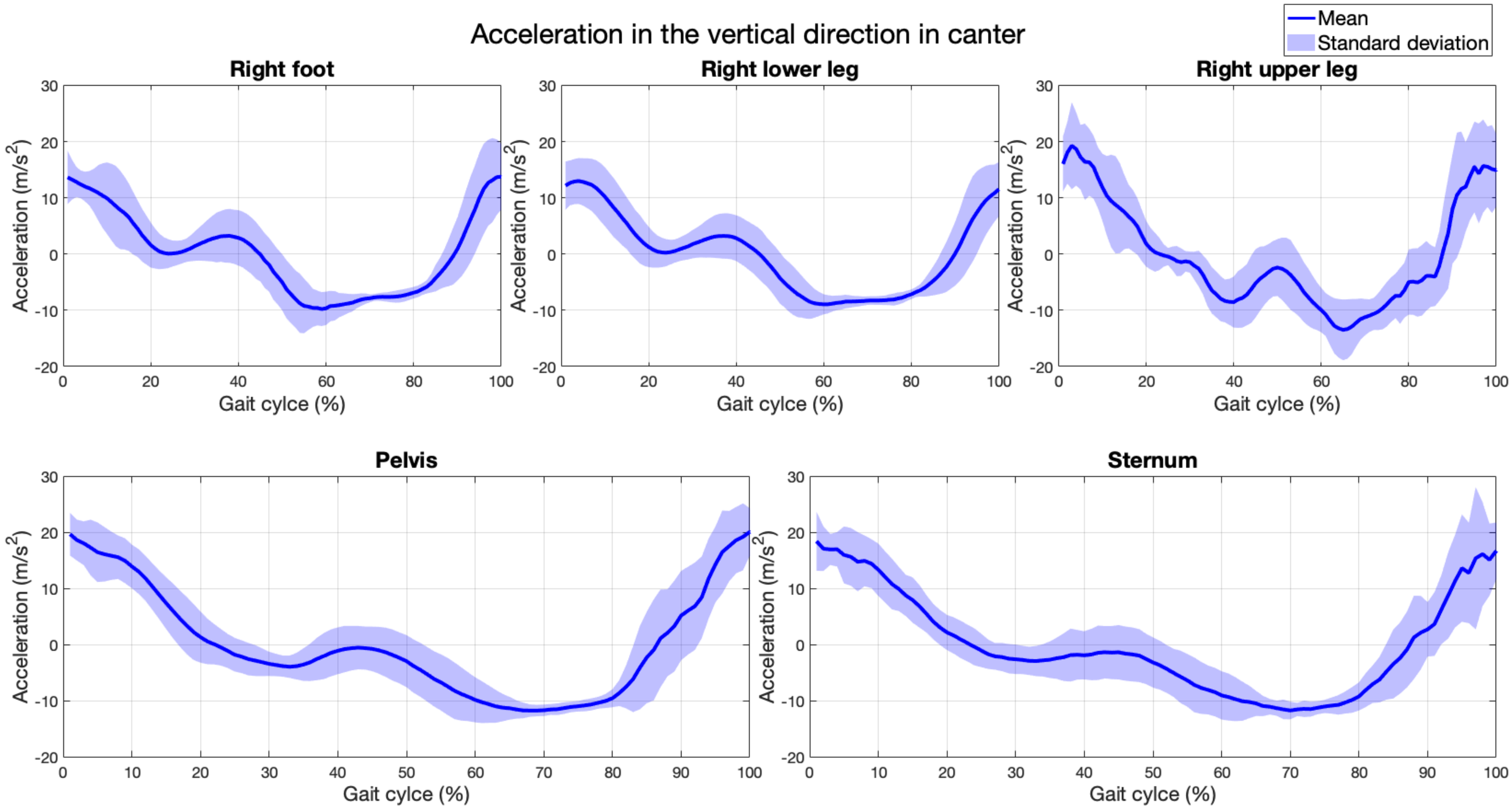


*Figure 6: Acceleration in the vertical direction in canter for the different segments.*

*Table 3: Peak acceleration values (in m/s$^2$) for sitting trot and canter, represented as mean $\pm$ standard deviation. RFO: right foot, RLL: right lower leg, RUL: right upper leg, PEL: pelvis, STE: sternum, S: conditions in which the participants were riding with stirrups, WS: conditions in which the participants were riding without stirrups. A significance level of 0.05 was used, an asterisk is used to denote significant differences.*

| | | **Peak acceleration vertical direction (m/s$^2$)** | | | **Peak acceleration modulus (m/s$^2$)** | | |
|---|---|---|---|---|---|---|---|
| | | S | WS | p-value | S | WS | p-value |
| **Sitting trot** | RFO | 17.3 $\pm$ 2.8 | 20.4 $\pm$ 2.8 | **0.023*** | 24.7 $\pm$ 6.6 | 27.1 $\pm$ 4.0 | 0.332 |
| | RLL | 12.9 $\pm$ 1.7 | 17.3 $\pm$ 2.0 | **<0.001*** | 17.3 $\pm$ 2.7 | 20.7 $\pm$ 3.0 | **0.015*** |
| | RUL | 33.3 $\pm$ 9.4 | 39.4 $\pm$ 10.0 | 0.177 | 54.5 $\pm$ 14.9 | 62.4 $\pm$ 21.0 | 0.347 |
| | PEL | 28.0 $\pm$ 5.6 | 33.3 $\pm$ 10.7 | 0.185 | 33.6 $\pm$ 8.2 | 40.4 $\pm$ 13.4 | 0.189 |
| | STE | 34.2 $\pm$ 5.5 | 38.2 $\pm$ 10.1 | 0.298 | 45.5 $\pm$ 12.7 | 52.2 $\pm$ 18.4 | 0.356 |
| **Canter** | RFO | 18.5 $\pm$ 3.1 | 18.2 $\pm$ 5.1 | 0.883 | 25.0 $\pm$ 3.7 | 26.0 $\pm$ 4.9 | 0.625 |
| | RLL | 16.8 $\pm$ 2.6 | 20.3 $\pm$ 5.1 | 0.067 | 21.9 $\pm$ 2.5 | 24.1 $\pm$ 4.8 | 0.207 |
| | RUL | 27.1 $\pm$ 6.1 | 35.2 $\pm$ 12.3 | 0.078 | 57.3 $\pm$ 19.4 | 74.4 $\pm$ 31.3 | 0.159 |
| | PEL | 22.1 $\pm$ 4.8 | 27.6 $\pm$ 9.4 | 0.118 | 33.2 $\pm$ 11.1 | 45.5 $\pm$ 25.9 | 0.183 |
| | STE | 22.9 $\pm$ 8.9 | 26.9 $\pm$ 11.3 | 0.387 | 31.4 $\pm$ 12.9 | 38.4 $\pm$ 21.4 | 0.385 |

*Table 4: Area values (in m/s) for sitting trot and canter, represented as mean $\pm$ standard deviation. RFO: right foot, RLL: right lower leg, RUL: right upper leg, PEL: pelvis, STE: sternum, S: conditions in which the participants were riding with stirrups, WS: conditions in which the participants were riding without stirrups. A significance level of 0.05 was used, an asterisk is used to denote significant differences.*

| | | **Area vertical direction (m/s)** | | | **Area modulus (m/s)** | | |
|---|---|---|---|---|---|---|---|
| | | S | WS | p-value | S | WS | p-value |
| **Sitting trot** | RFO | 3.9 ± 0.7 | 4.1 ± 0.7 | 0.468 | 11.6 ± 2.0 | 12.1 ± 1.8 | 0.560 |
| | RLL | 2.8 ± 0.3 | 3.1 ± 0.3 | **0.033*** | 8.7 ± 1.2 | 9.6 ± 1.3 | 0.110 |
| | RUL | 5.0 ± 0.8 | 5.3 ± 1.0 | 0.608 | 16.0 ± 3.0 | 16.7 ± 3.5 | 0.676 |
| | PEL | 4.2 ± 0.4 | 4.4 ± 0.4 | 0.466 | 10.8 ± 1.4 | 11.2 ± 1.7 | 0.526 |
| | STE | 4.3 ± 0.5 | 4.4 ± 0.7 | 0.764 | 11.1 ± 2.1 | 11.8 ± 2.7 | 0.511 |
| **Canter** | RFO | 2.5 ± 0.3 | 2.4 ± 0.3 | 0.479 | 7.5 ± 1.0 | 7.3 ± 0.8 | 0.593 |
| | RLL | 2.3 ± 0.3 | 2.4 ± 0.3 | 0.425 | 6.6 ± 0.9 | 6.7 ± 1.0 | 0.849 |
| | RUL | 3.6 ± 0.6 | 3.7 ± 0.7 | 0.706 | 13.3 ± 2.8 | 13.9 ± 3.1 | 0.663 |
| | PEL | 3.1 ± 0.4 | 3.4 ± 0.4 | 0.071 | 8.2 ± 1.1 | 9.2 ± 1.6 | 0.095 |
| | STE | 2.8 ± 0.3 | 3.0 ± 0.3 | 0.111 | 7.0 ± 1.0 | 7.9 ± 1.3 | 0.114 |

Repeated-measures ANOVA was used to detect differences between the conditions with and without stirrups (see Table 3 and Table 4). It was found that in sitting trot, the peak acceleration in the right foot was significantly higher for the condition without stirrups when looking at the acceleration in the vertical direction (p = 0.023). Furthermore, the peak acceleration in the right lower leg was significantly higher in the condition without stirrups for both the vertical direction (p = < 0.001) and the modulus (p = 0.015). Looking at the area under the curve, the only significant difference was a larger area under the curve for the right lower leg in the condition without stirrups when only acceleration in the vertical directions were considered (p = 0.033). In canter, no significant differences were found. The shock attenuation in the lower extremities as calculated by looking at the peak acceleration in the vertical direction was significantly lower for the condition without stirrups in sitting trot (p = 0.02). The shock attenuation in the lower extremities as calculated by the area under the curve considering only the acceleration in the vertical direction was significantly lower for the condition without stirrups in canter (p = 0.009). No significant differences were found for the shock attenuation in the torso between the condition with and without stirrups. The shock attenuation from pelvis to sternum in sitting trot, however, was significantly lower when calculating it based on the area compared to the peaks for both the vertical direction (p = 0.003, paired samples t-test) and the modulus (p = 0.003, paired samples t-test).

Due to the unexpected findings of a significantly higher superior acceleration in the trunk, a case study with a different measurement system was performed. This case study is presented in section 3.1.

## 3.1 A case study to investigate the effects of MVN Analyze' biomechanical model on pelvis and sternum accelerations

A case study was performed to better comprehend the results that are presented in this thesis. In this case study, a different measurement system was used in which it is possible to access the raw data. The measurement system and software used in the general study uses a biomechanical model of which some details are unknown (22). The software uses sensor fusion algorithms that alter the raw data. The goal of this case study was to investigate whether the higher sternum acceleration found in the general study might be due to the biomechanical model used in the MVN Analyze software.

*Methods*

*Participant*

The pilot study was performed on one female horseback rider (age: 26 years, height: 185 cm, weight: 75 kg, BMI: 21.9, riding experience: 11 years, riding level: Dutch ZZ-licht level) and her horse (Dutch warmblood (KWPN), age: 7 years, height at withers: 179 cm). This rider has had multiple surgeries on her left ankle and right shoulder. Nevertheless, the case study was performed on this participant as it was mainly intended as a means to better comprehend the results presented in this thesis and not to investigate biomechanical relationships.

*Experimental design*

In the case study, wireless IMUs were used. Five Xsens DOT sensors (Xsens Technologies BV, Enschede, The Netherlands) were taped to the skin of the participant at the same locations where they would be in the Lycra suit. As only five sensors were available, the sensors were placed on the right foot, right lower leg, right upper leg, pelvis and sternum. The sensors were synchronized using the Xsens DOT app (version 2021.0). The 'Recording (Offline mode)' was used, resulting in a sample rate of 120 Hz.

*Procedure*

Data collection took place in an indoor riding arena of 20 by 40 meters. The participant provided her horse with a standard warm-up. After the warm-up, the participant performed the same protocol as described in section 2.2.2.

*Analysis*

Data was analyzed in the sensor frame, a right-handed coordinate Cartesian system in which the x-axis of the pelvis and sternum sensor are approximately equal to the z-axis of the global reference frame used in the general study. This holds true if the pelvis and sternum remain in an upright position and do not rotate with respect to each other. Data was analyzed in MATLAB 2020a. The measurements of the pelvis and sternum were analyzed in the approximately vertical direction (the x-axis). The data from this pilot was only visually inspected to validate the patterns and values found in the general study.

*Results*

The movement patterns and peak values of the data collected from the pelvis and sternum sensors were compared to the results of the general study. Similar acceleration patterns and peak values were observed, with higher peak accelerations in the sternum compared to the pelvis. Furthermore, it was noted that there are small peaks present when the acceleration is at its highest point which coincide with a forward tilting rotational movement of the pelvis (anterior pelvic tilt) that is not fully transferred to the sternum.

To aid the visual inspection of the data, the difference in acceleration along the x-axis and the difference in angular velocity around the y-axis were plotted. These were defined as follows:

$$\textit{Difference in acceleration} = \textit{sternum acceleration} - \textit{pelvis acceleration} \quad (4)$$

$$\textit{Difference in angular velocity} = \textit{angular velocity pelvis} - \textit{angular velocity sternum} \quad (5)$$

Figure 7 and Figure 8 show the difference in acceleration and the difference in angular velocity for a representative selection of gait cycles for sitting trot and canter respectively.

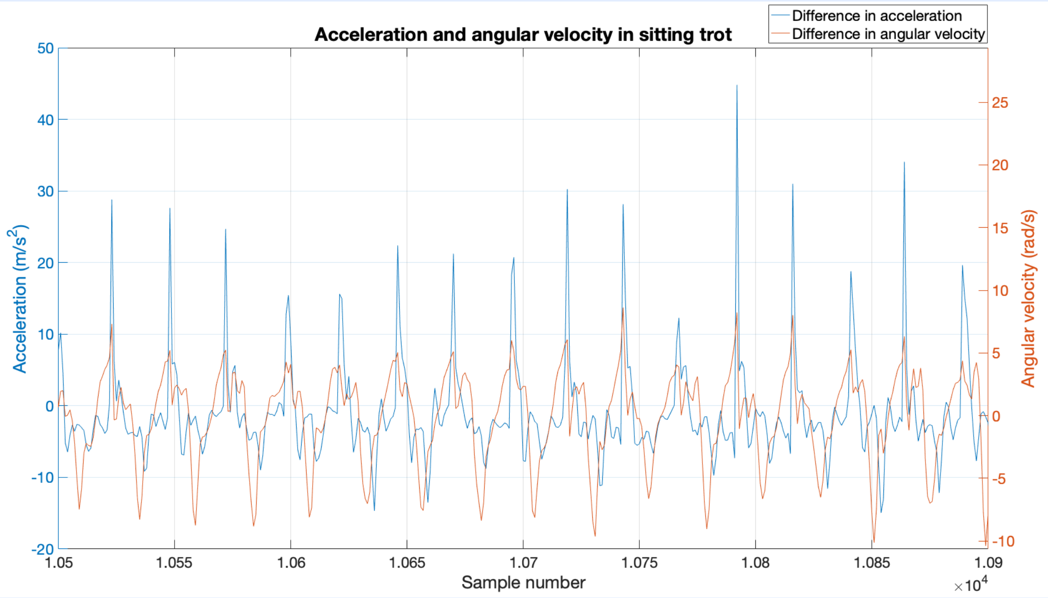


*Figure 7: Acceleration and angular velocities in sitting trot. The difference in acceleration in the approximately vertical direction between the sternum and pelvis is displayed on the left axis (in blue). The difference in angular velocity around the y-axis between the pelvis and sternum is displayed on the right axis (in red).*

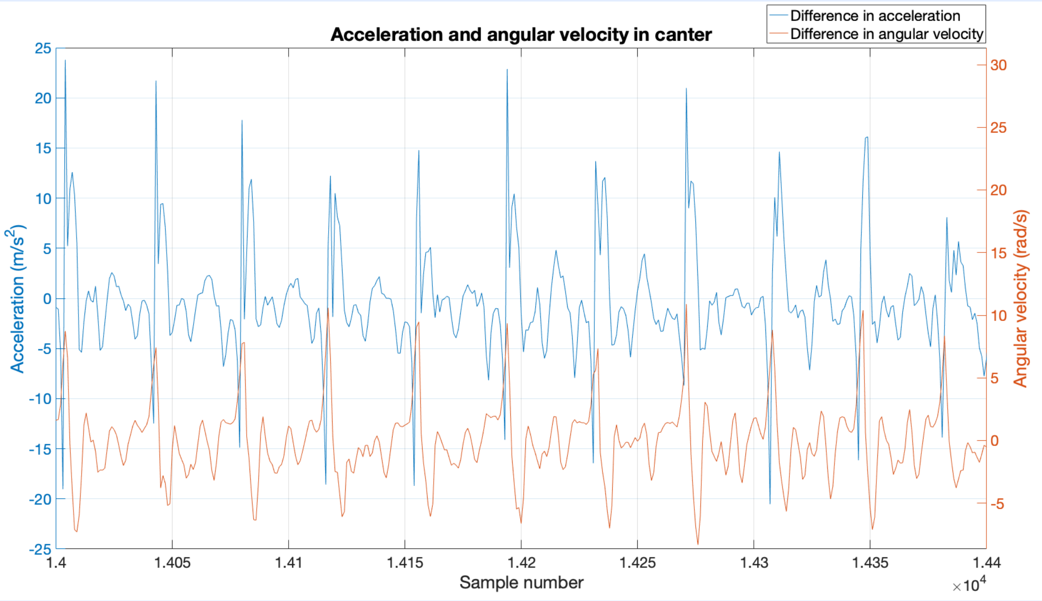


*Figure 8: Acceleration and angular velocities in canter. The difference in acceleration in the approximately vertical direction between the sternum and pelvis is displayed on the left axis (in blue). The difference in angular velocity around the y-axis between the pelvis and sternum is displayed on the right axis (in red).*

# 4. DISCUSSION

This study has shown that IMUs can be used to objectify biomechanical parameters in horseback riders without the need for an additional sensor on the horse itself. Hip angles were used to detect the gait cycle in the different gaits. We hypothesized that gait cycle detection could be done based on pelvic angles. However, in the current study, hip angles showed a more clearly defined movement pattern, making it easier to identify the gait cycle based on hip angles. This technique works well for sitting trot and canter. In walk, the movement patterns were not always clear. For some participants, using pelvic angles instead of hip angles improved gait cycle detection. However, gait cycle detection in walk remained difficult for some participants, requiring manual selection of the gait cycles by visual inspection the data. Thus, the gait cycle of a horse can be automatically detected by using hip angles of the rider in sitting trot and canter. Gait detection also works in walk, but might require more manual labor.

Clear acceleration patterns were visible in sitting trot and canter. The low absolute values of the acceleration that are measured in walk make it difficult to properly analyze this data and form conclusions or recommendations. In sitting trot, it was found that the acceleration in the vertical direction of the sternum was significantly higher than the acceleration in the vertical direction of the pelvis. This is not in line with expectations, as shock attenuation is normally aimed at keeping proximal accelerations minimal. There is no additional external force applied above the pelvis which could explain the increase in acceleration in the vertical direction. Results from the case study with the Xsens DOT sensors suggest that there might be a rotational component involved. The sudden forward rotation of the pelvis from a leaned-back position could create a linear acceleration in the torso. This could potentially explain the higher acceleration values in the sternum found in this study. Visual inspection of the accelerations in the vertical direction and the modulus in sitting trot and canter suggests that the vertical acceleration is dominant in sitting trot, whereas the other planes of motion might have a larger influence in canter. This is congruent with the movement of the rider on a horse.

In this study, there is a decreased magnitude of the negative shock attenuation from pelvis to sternum in sitting trot when looking at the area instead of the peak value. This could indicate that the total impact on both segments might be similar, but that the impact is spread out over a larger portion of the gait cycle for the pelvis compared to the sternum. On a group level, significant differences were found in the shock attenuation with and without stirrups from the right foot to the right lower leg. Riding without stirrups led to a significantly higher acceleration in the right lower leg and a significantly lower shock attenuation from the right foot to the right lower leg. This suggests that, when riding with stirrups, the ankle is of importance with regards to shock attenuation. No other significant differences were found between absolute acceleration values or shock attenuation values with and without stirrups. Riding without stirrups does not seem to influence the accelerations in the pelvis. Thus, even though there is shock attenuation between the foot and lower leg, the stirrups do not seem to be of importance with regards to the impacts on the trunk of the rider.

## 4.1 Implications for CLBP

With regards to CLBP, it was hypothesized that the shock attenuation in the vertical direction could give information about possible compression forces in the spinal column. Although a negative shock attenuation was found from pelvis to sternum in sitting trot, the variation between participants was too large to make any robust conclusions. Shock attenuation became more negative when looking at the modulus instead of the vertical direction, although the variance also increased. This could indicate that the spinal column is also exposed to shear or rotational forces. These findings could be of interest with regards to the development of CLBP.

An increase in acceleration was found in the torso, with significantly higher acceleration values being found in the more superior segment. Based on the results from the case study, it is theorized that this phenomenon could be explained by a rotational component from the pelvis that is not fully transferred to the sternum. This suggests that core stability and flexibility is of importance in horseback riders. Core stability might be of importance to stabilize the trunk during this rotational movement, which could imply that riders with a limited core stability have a higher risk of developing CLBP. Furthermore, a certain amount of flexibility needs to be present in the spinal column to attenuate the rotational movement. Therefore, horseback riders who have limited flexibility in the spine might also have a higher risk of developing CLBP. Obesity or an increased age could thus be potential risk factors for the development of CLBP in horseback riders. On the other hand, training regimes focused on improving core stability could reduce the risk of CLBP.

## 4.2 Limitations and recommendations

In all measurements, the accelerations of the right upper leg are larger than expected, with a less smooth pattern. This is especially visible when looking at the modulus of the acceleration. The upper leg is in contact with the saddle over the entire segment length. A potential explanation for the large and unsteady acceleration pattern is that the contact between the upper leg and the saddle is not constant, thereby leading to peaks in the acceleration pattern. It is also possible that tissue artifacts or muscle contraction in the upper leg influenced the acceleration data presented here. Filtering techniques or frequency analysis could be applied to improve the data quality of the upper leg.

In walk, the absolute acceleration values were low and gait cycle detection was not as easy compared to sitting trot and canter. This is most likely due to the low impacts that a rider experiences during walk. Walk is the slowest of the three paces and is a 4-beat movement, meaning that the movements of one gait cycle are evenly distributed over the separate movements of the 4 legs of the horse (see Figure 4). Therefore, the impacts are very low and the movements that the rider makes are also very small. Since the movements are very small, the movement pattern in joint angles is not as clearly visible as in sitting trot and canter. A potential solution could be to place an additional sensor on the trunk to see if the relative angle between those two sensors would give a clear movement pattern that allows for easy gait cycle detection. However, even if gait cycle detection in walk is improved, the absolute acceleration values will remain very low. IMUs with a smaller margin of error with regards to the acceleration are most likely needed to detect clear acceleration patterns in walk.

Large variety was found in the shock attenuation values calculated. This suggests that the shock attenuation strategies differ between riders. A personalized approach to the study of biomechanics of horseback riders is therefore warranted. Furthermore, the method by which shock attenuation is calculated (vertical direction or modulus, peak or area) seems to influence the values found. This could indicate that the shock attenuation strategies differ per segment. A better understanding of what the area under the curve represents could be beneficial to explain these differences.

The data used in the general study has been processed in MVN Analyze, in which a biomechanical model is used of which some details are unknown. The case study with the Xsens DOT sensors provided less processed data. Although the data from the case study is in a sensor frame, it can be assumed that the x-axis of the sensor frame is approximately equal to the vertical direction in the general study for the pelvis and sternum sensors. There is some rotational movement between the pelvis and sternum, but this movement is considered to be small enough to allow for a visual comparison between the data from the case study and the general study. Visual comparison of the data suggests that the processing pipeline in MVN Analyze did not significantly alter the characteristics of the data. However, data collected during the case study showed smaller, presumably impact, peaks at the points where the acceleration increased. This was not visible in the data obtained during the general study, most likely due to a filtering process in MVN Analyze. If there are indeed multiple smaller impact peaks present, as the data from the case study suggests, then this could have implications with regards to CLBP. It could be hypothesized, for example, that the spinal column has more difficulty with attenuating multiple repeated impacts compared to one longer impact of the same magnitude. Therefore, it might be interesting to use sensors which are able to provide raw, unprocessed data in future studies.

Furthermore, the data from the case study suggests that there might be a rotational component in the pelvis which leads to a higher linear acceleration in the sternum. If a rotational component is indeed involved, this highlights the importance of core stability in horseback riders to prevent (back) injuries. It was difficult to visualize or correct for this rotational component in the data from the general study, since the horse and rider changed direction multiple times with the current study protocol. Therefore, the axes of the global reference frame also changed direction multiple times. A suggestion could be to rotate the general reference frame such that the x-axis continuously points in the direction of motion. This would allow for the visualization and quantification of the rotational component. This is preferable to using the sensor frame, since the z-axis of the sensor frame does not always coincide with the vertical direction.

The study population used in this research project comprised only of healthy females of a relatively young age. None of the participants in the current study had a history of CLBP. Given the high prevalence of CLBP amongst horseback riders, this could suggest that the results shown here are representative of a good riding posture and riding technique. It would therefore be interesting to include participants with CLBP in future research to allow for the comparison of (peak) acceleration and shock attenuation with healthy controls. The current data set could also be expanded by including participants of different age categories and by including male horseback riders. Given the different body composition of males and females, it is possible that different patterns are observed in male riders. For older riders, the flexibility of the tissues and joints might be decreased, leading to less efficient shock attenuation.

Furthermore, the study protocol used here was designed to limit the number of aids that a rider needs to give to the horse. During a normal training, however, riders a constantly giving their horses aids. Aids can be given via the lower extremities, the trunk and the upper extremities. Thus, during a normal training session, there will be many active contributions of the rider to the movements of the segments. This could also be investigated further by measured riders during a normal training, where measurements in rising trot are also included. Future research could also include the upper extremities to give a more complete overview of the accelerations and forces that are present during horseback riding.

The current study does not differentiate between straight lines and curves during riding. The curves could influence the measured parameters, although this was not directly visible upon visual exploration of the data. The protocol of the current study deliberately included both straight lines and curves, since this is more in accordance with the normal riding situation. Several parameters were investigated to see if detection of straight lines was possible. No straightforward method was found, although the angular velocity of the pelvis seems to indicate the straight lines and curves for some participants. Future studies could include straight lines and curves as different parts of the protocol so that distinctions can be made, and potential differences can be explored.

Using video analysis or an additional sensor on the horse can also be interesting to see if gait events, such as the landing of a hoof, can be linked to patterns observed in the rider. The relationship between the acceleration peaks in the rider and the movement pattern of the horse is not as straightforward as it is in, for example, walking or running. The different paces of the horse are 2-, 3- and 4-beat (trot, canter and walk respectively), which could have implications for the acceleration patterns in the rider. It might also be interesting to perform measurements on different riders on the same horse. The differences in movements between horses can create differences in, for example, the absolute value of the acceleration in the rider. The influence of the horse would be eliminated by letting all participants perform the protocol on the same horse.

# 5. CONCLUSIONS

An 8 IMU set-up can be used to objectify (peak) acceleration and shock attenuation in healthy horseback riders in sitting trot and canter. With the method proposed in this study, there is no need for an additional sensor on the horse to define the gait cycle. Gait detection is also possible in walk, although it might require more manual labor. Acceleration values in walk were too low to detect clear patterns and peaks.

Contrary to the hypothesis, significantly higher peak acceleration values of the sternum compared to the pelvis were found in sitting trot. This indicates that the shock might be increased, rather than attenuated. Data from the case study indicated that a rotational component might be involved in the torso. It is hypothesized that a rotational movement of the pelvis leads to a linear acceleration in the sternum. This could be of interest with regards to CLBP and highlights the importance of core stability for horseback riders.

## 6. ACKNOWLEDGEMENTS

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# APPENDIX

## **Appendix A:** Accelerations in walk

Figures 9 and 10 show the acceleration in walk for the vertical direction, for the conditions with and without stirrups respectively. Figures 11 and 12 show the modulus of the acceleration in walk for the conditions with and without stirrups, respectively.

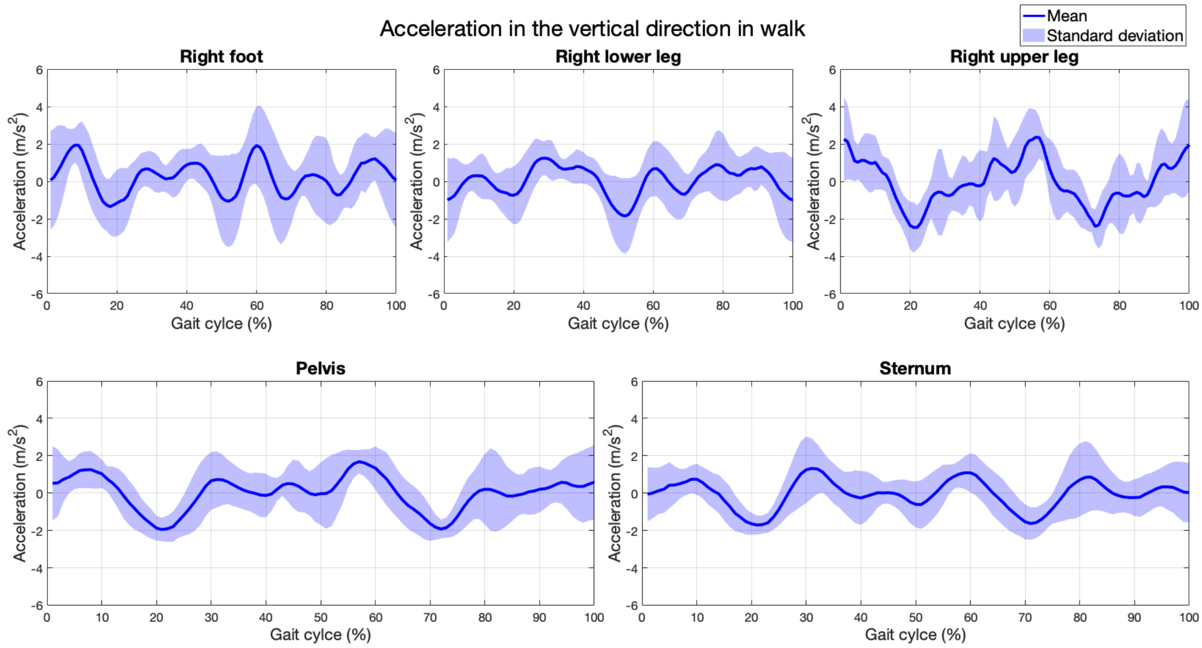


*Figure 9: Acceleration in the vertical direction in walk with stirrups for the different segments.*

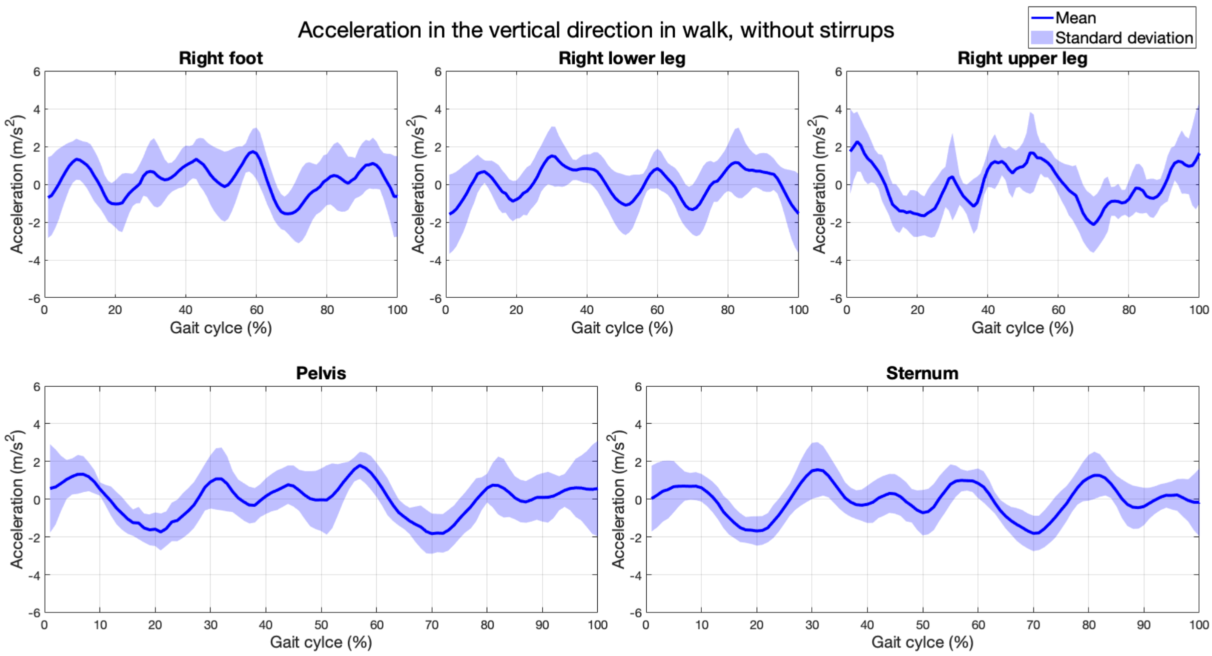


*Figure 10: Acceleration in the vertical direction in walk without stirrups for the different segments.*

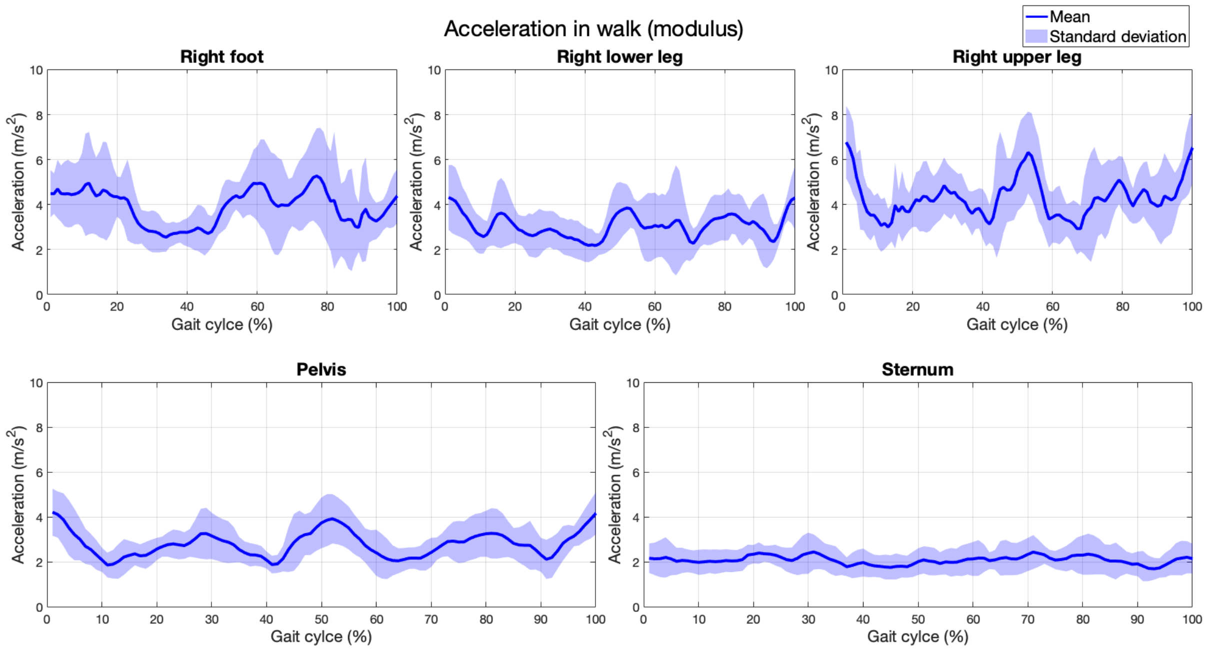


*Figure 11: Modulus of the acceleration in walk with stirrups for the different segments.*

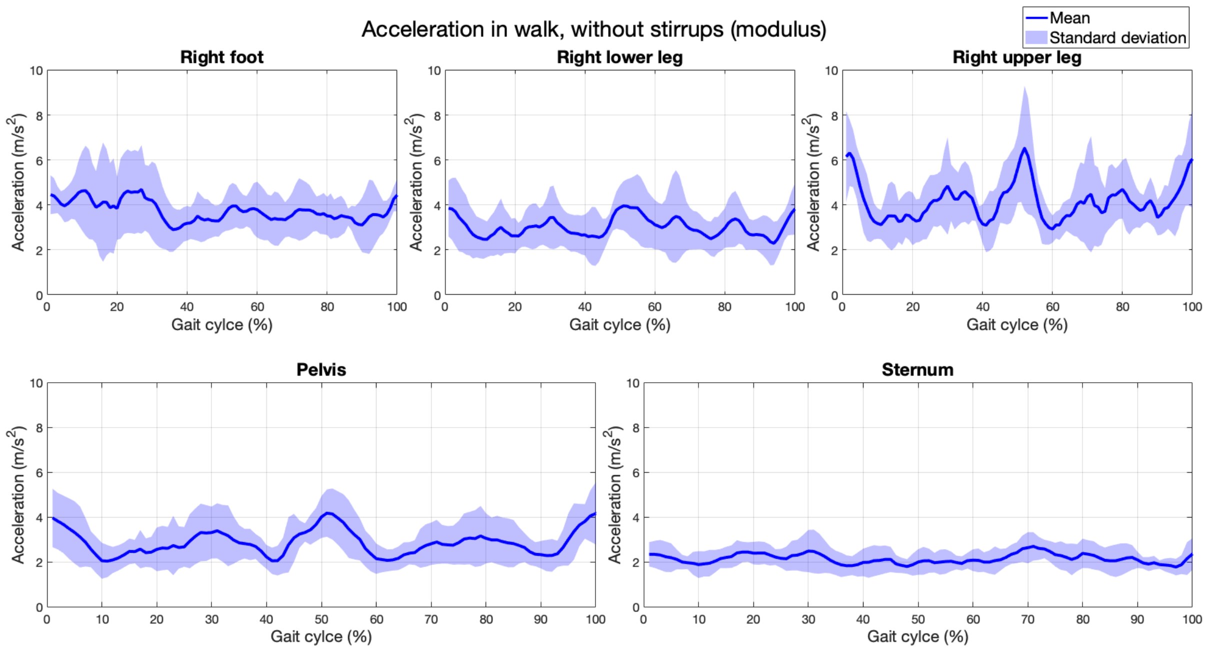


*Figure 12: Modulus of the acceleration in walk without stirrups for the different segments.*

## **Appendix B:** Accelerations in sitting trot

Figures 13 shows the acceleration in the vertical direction in sitting trot without stirrups. Figures 14 and 15 show the modulus of the acceleration in sitting trot with and without stirrups, respectively. Figure 16 shows the modulus of the acceleration in sitting trot without stirrups for the right upper leg.

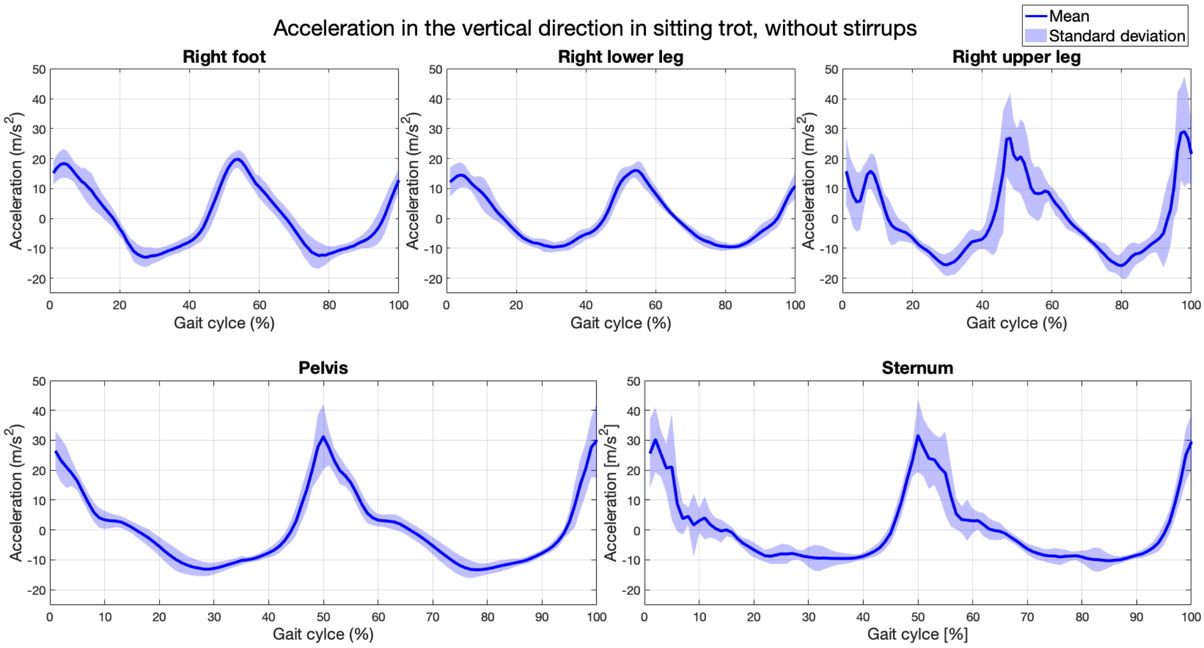


*Figure 13: Acceleration in the vertical direction in sitting trot without stirrups for the different segments.*

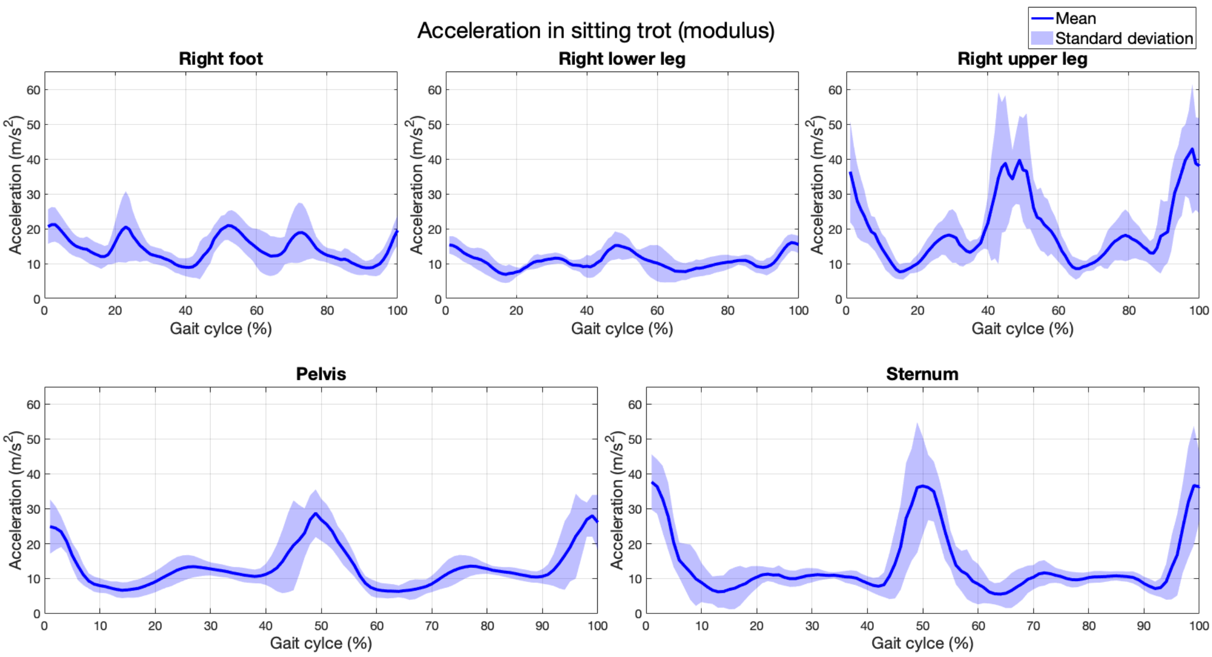


*Figure 14: Modulus of the acceleration in sitting trot with stirrups for the different segments*

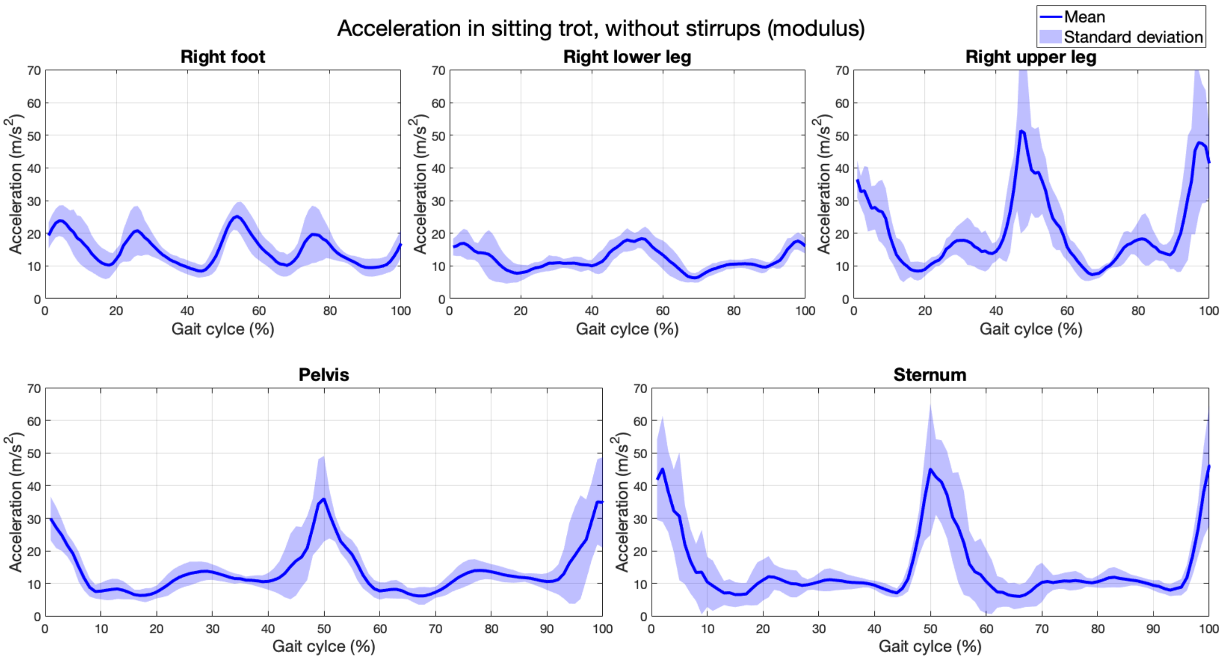


*Figure 15: Modulus of the acceleration in sitting trot without stirrups for the different segments. The standard deviation of the right upper leg falls outside of the acceleration range shown in this figure. Figure 16 fully shows the acceleration pattern of the right upper leg.*

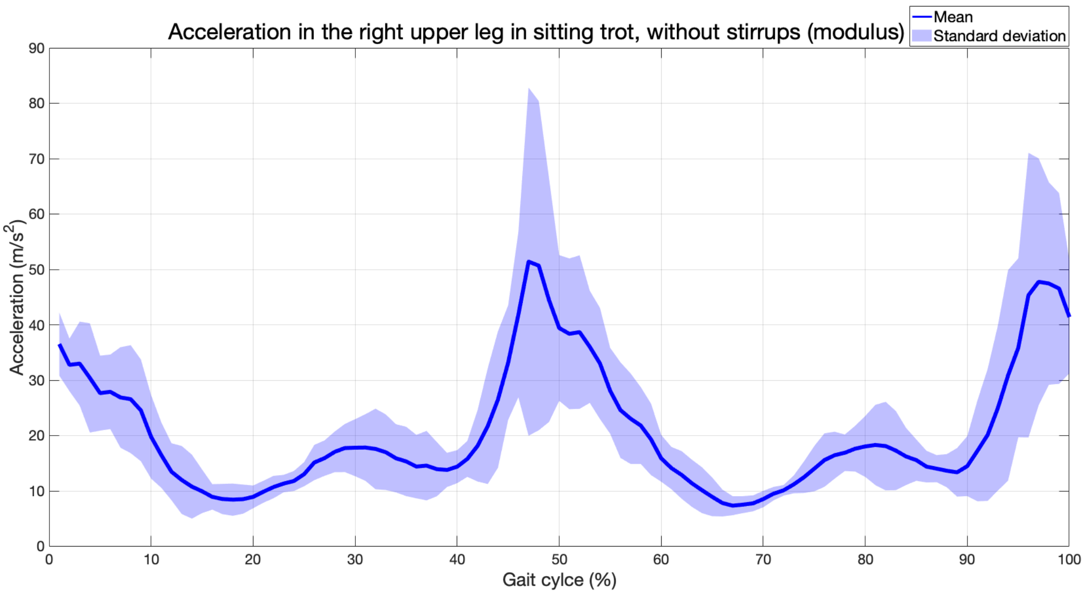


*Figure 16: Modulus of the acceleration in sitting trot without stirrups for the right upper leg.*

## **Appendix C:** Accelerations in canter

Figures 17 shows the acceleration in the vertical direction in canter without stirrups. Figures 18 and 19 show the modulus of the acceleration in canter with and without stirrups, respectively. Figure 20 shows the modulus of the acceleration canter with and without stirrups for the right upper leg.

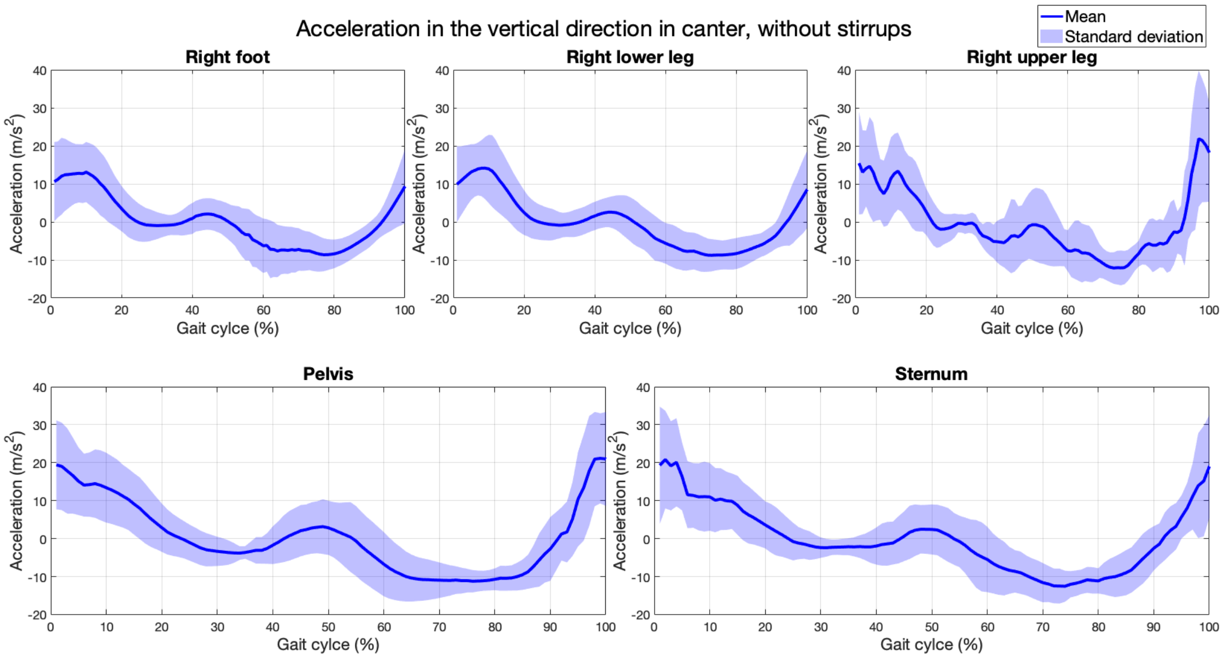


*Figure 17: Acceleration in the vertical direction in canter without stirrups for the different segments.*

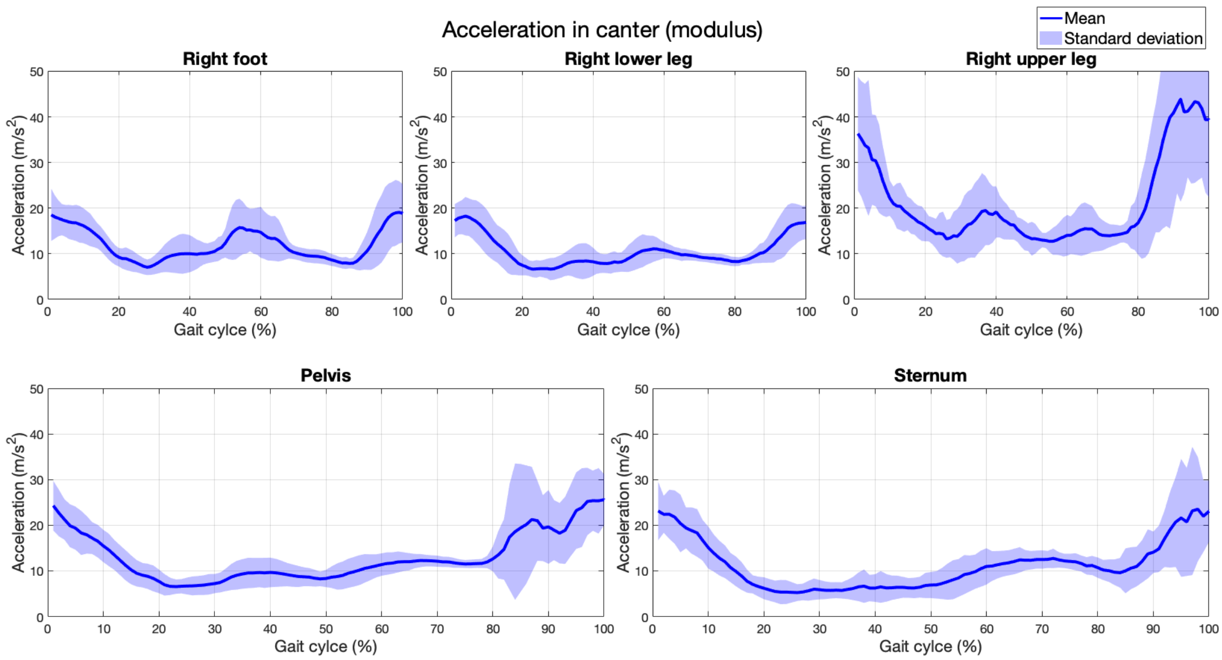


*Figure 18: Modulus of the acceleration in canter with stirrups for the different segments. The standard deviation of the right upper leg falls outside of the acceleration range shown in this figure. Figure 20 fully shows the acceleration pattern of the right upper leg.*

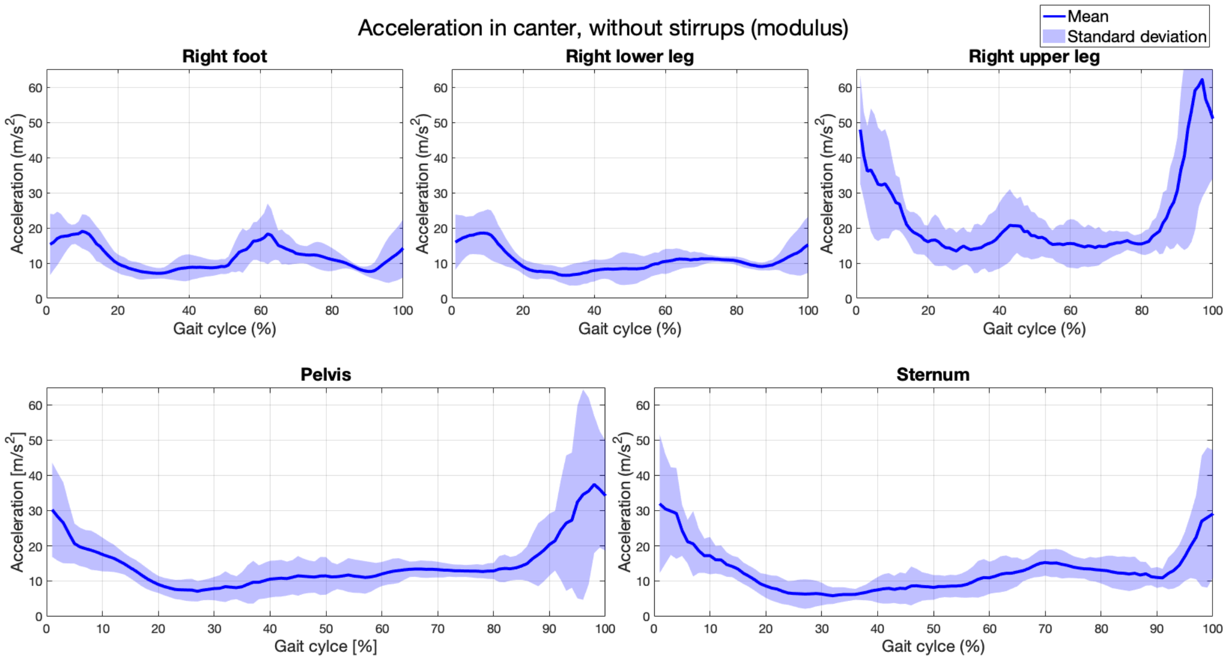


*Figure 19: Modulus of the acceleration in canter without stirrups for the different segments. The standard deviation of the right upper leg falls outside of the acceleration range shown in this figure. Figure 20 fully shows the acceleration pattern of the right upper leg.*

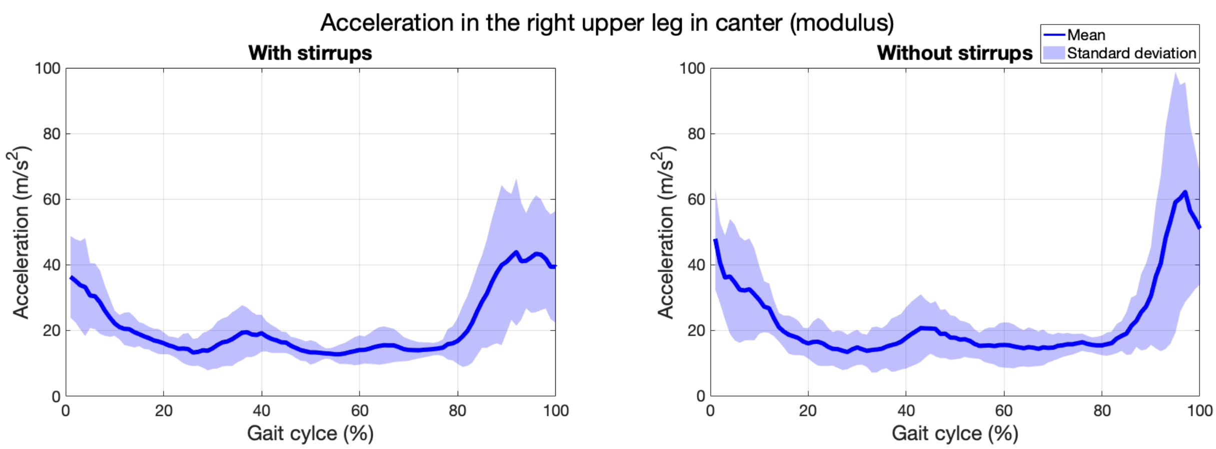


*Figure 20: Modulus of the acceleration in canter with and without stirrups for the right upper leg.*