A Biomechanical Study on the Hoof Impact at the Trot

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Abstract

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This thesis is a study of the hoof deceleration and the propagation of impact related shock waves through the structures of the distal limb in the beginning of the stance phase. The equine limb has an in-built shock absorbing capacity, with an anatomy that gives a successive disto-proximal loading that prolongs the time of loading uptake. The repetitive impulsive loading subjected to the limb following each hoof impact has been suggested to be an important factor in the mechanical stress that if sufficiently repeated may lead to subchondral bone damage.
The horses used in the study were Standardbred trotters. The deceleration pattern of fore- and hind hooves were recorded with accelerometers at different circumstances.
In the first study, horses trotted at slow speed over a force plate. with one accelerometer glued to the hoof wall, and one fixed to the third metacarpal bone, in order to record the transmission of impact transients. The initial impact peak was assumed to be negligible proximal to the fetlock joint. The length of the hoof braking seemed to influence on the attenuation of the following peak decelerations. In the following studies only hoof mounted accelerometers were used. In the second study the deceleration patterns of the fore- and hind hooves were compared over the force plate on a sand surface at slow trot. In the third study the hoof deceleration was compared between two different surfaces. The forth study was a field study on the hoof decelerations at different speeds (4.7-12.7 ms^{-1}) on a training race track. The hoof deceleration is suggested to be divided into two parts. The first is characterized by large impact peak that is mainly attenuated distal to the fetlock joint. The second part is characterized by the onset of loading of the fetlock joint and the following rapid movements of distal bone segments, and the interaction between the ground surface and the hoof. The loading rate and the magnitude of the horizontal peak decelerations increase with speed and with a higher friction between the ground and hoof. The results improve the knowledge of the shock absorbing capacity of the equine limb and the mechanisms behind indicators of mechanical stress to the limb.

Keywords: accelerometer, attenuation, horse, hoof braking, kinetics, kinematics, locomotion, mechanical stress

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Appendix

Papers I-IV

The present thesis is based on the following papers, which will be referred to by their Roman numerals:

I. Gustås, P., Johnston, C., Roepstorff, L. & Drevemo, S. 2001. *In vivo* transmission of impact shock waves in the distal forelimb of the horse. *Equine veterinary journal supplement 33*, 11-15.

II. Gustås, P., Johnston, C., Roepstorff, L., Drevemo, S. & Lanshammar, H. 2004. Relationships between forelimb and hind limb ground reaction force and hoof deceleration patterns in trotting horses. *Equine veterinary journal 36*, 737-742.

III. Gustås, P., Johnston, C. & Drevemo, S. Ground reaction forces and hoof deceleration patterns on two different surfaces at the trot. (Manuscript).

IV. Gustås, P., Johnston, C., Hedenström, U., Roepstorff, L. & Drevemo, S. A field study on hoof deceleration at impact in Standardbred trotters at various speeds. (Manuscript).

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Introduction

Background

The rapid and repeated loading of the limb at hoof impact have been speculated to be associated with the development of lameness due to musculoskeletal injuries in horses in training (Hjertén & Drevemo, 1994; Merkens & Schamhardt, 1994).
These suggestions stem from the experiments by Radin *et al.* (1973), in which rabbits subjected to repetitive impulse loading, demonstrated stiffening of the subchondral bone. Further experiments with animal models (Serink *et al.* 1977; Dekel & Weissman, 1978) supported these observations. Later, Radin *et al.* (1991) suggested the large peak decelerations and a high loading rate following impact to be indicators of mechanical stress. However, the concepts of impact forces and the paradigms of their attenuation are not well understood in relation to the etiology of injuries and complex mechanisms may be responsible for injury development.

It is well known that many Standardbred trotters suffer from injuries to the structures of the locomotor system during their careers. A large part of these injuries affect the carpus and tarsus and structures distal to these joints. Several studies describe different sources of mechanical stress as underlying causes to excessive wear and overload of the tissues of the locomotor system leading to training related injuries (James & Jones, 1990; Wosk & Voloshin, 1981). The mechanisms behind the development of tissue damage possibly due to mechanical stress are not described in detail in the horse, but it is assumed that the exposure to external forces is an important factor (Norrdin *et al.* 1998).
One main option with training is to increase the load on the body successevely, in order to encourage tissue adaptation to excessive load, and thereby increase the strength for improved performance (Nigg, 1986). For the training of the horse, it is therefore of great importance to learn more about the mechanisms behind mechanical stress to the limb. This knowledge is a major prerequisite for decisions how to adjust the mode of training in order to initiate, maintain or improve the mechanisms of remodelling and reconstruction of tissues to better withstand higher load without inducing injury.

Equine biomechanics of hoof impact and the early stance phase

High speed kinematic studies of Standardbred trotters in the racetrack have demonstrated irregularities in the movements of the hoof following first contact (Fredricson *et al.* 1972). Despite the large variation of the movement pattern at the hoof in individuals, the more proximal segments of the limb showed a largely repetitive movement pattern. From these findings it was suggested that horses have an effective mechanism of damping impact loading. It was further found that the Standardbreds in general land heel first (Fredricson and Drevemo, 1972) and often towards the lateral side of the hoof (Dalin *et al.* 1973).
The ground reaction force patterns from measurements on a force plate at walk, trot and canter were presented by Pratt & O'Connor (1976). They observed the impact peak following the initial ground contact.

The impact peaks of the canter were discussed, however in the trot, these peaks were not mentioned in the text but were visualized in a figure. Also the role of the horizontal force in the deceleration of the hoof was mentioned. Further studies followed on the ground reaction force characteristics of the horse (Ueda *et al.* 1981; Niki *et al.* 1982 and 1984; Goodship *et al.* 1983; Merkens *et al.* 1985, 1994; Hjertén and Drevemo, 1987; Morris and Seeherman, 1987; Seeherman *et al.* 1987). In force-time curves of these studies an impact peak is not always present, but there is a clearly visible steep increase in the vertical force, indicating a higher loading rate during the time period following first contact.

A low effective mass following first contact and a slow increase in load during the following milliseconds were described by Hjertén & Drevemo (1994).
In accordance with the early suggestions, the equine limb was described as an effective shock absorber (Hjertén & Drevemo, 1993, 1994; Hjertén *et al.* 1994), due to the successive loading in its longitudinal direction from the hoof and upwards. They concluded that the mode of loading prolongs the impact time, thus resulting in lower impact forces compared to a stiff limb.

Detailed descriptions of the fore- and hind limb kinematics at slow trot have been presented by van Weeren *et al.* (1993); Back *et al.* (1995a, 1995b, 1995c), and at high speed by Johnston *et al.* (1995, 1996).

Accelerometer studies

The use of accelerometers gives the possibility to study the direct effect of ground reaction forces on the movement patterns of the hoof. Barrey *et al.* (1991) and Benoit *et al.* (1993) used accelerometers to distinguish and compare the different behaviours of the hoof on different ground surfaces and with different shoeing.
Hjertén & Drevemo (1994) presented accelerometer measurements combined with measurements from a force shoe (Björck, 1958) and high speed cinematography. Merkens & Schamhardt (1994) used an accelerometer to determine first contact.
The technique has been utilized for the study of transmission of impact related shock waves through the structures of the distal limb, both *in vivo* and *in vitro*.
Dyhre-Poulsen *et al.* (1994) used a bone mounted accelerometer to study the impact transients at the level of the proximal phalanx and the transmission was discussed in relation to the results presented by Benoit *et al.* (1993).
The transmission of impact peak characteristics both with respect to amplitude and frequency attenuation has been studied in different *in vitro* models by Lanovaz *et al.* (1998) and Willemen *et al.* (1999). Willemen *et al.* (1999) also presented an *in vivo* study of the propagation of impact related shockwaves through the distal bone segments. These studies showed amplitude attenuation between every bone segment. The frequency attenuation was to the greatest extent within the soft tissues of the hoof, which was suggested to function as a low-pass filter.

Impact mechanics during locomotion in general

In classical impact mechanics the word impact refers to the collision between two bodies. Forces generated at the impact are characterized to be large and act over a short period of time (Meriam & Kraige, 1987). Nigg *et al.* (1995) concluded that the landing of the foot could be described as a collision between two objects and therefore called an impact. Consequently, the term impact force is used for the force of short duration that arises due to the collision. An impulsive force is described as a force of relative short duration and relative large magnitude. Impact forces in running are characterized by a maximum, impact peak, reaching its maximum within 50 ms after first contact. Loading rate is the derivative of the force as a function of time (Hennig & Lafortune, 1991).
Similar terminology is also used in studies of hoof impact in equine biomechanics (Merkens *et al.* 1985; Hjertén and Drevemo, 1993; Hjertén *et al.* 1994).

Impact force propagation

The following short explanation of the different damping mechanisms involved in the impact force propagation is referring to a summary published by Nigg *et al.* (1995).
If the bone segments of the limb of the horse are simplified to a set of rigid segments connected by joints, aligned so that the angle between adjacent segments is 180° at impact, a force will be generated at the point of application. The magnitude of the impact force will be dependent on the kinetic energy of the system. Assuming linear motion the kinetic energy is dependent on the mass and vertical velocity of the system.
At impact a state of disequilibrium exists in the segment, and a longitudinal stress wave is propagated. If it is assumed that the segments are not elastic, they will move with the velocity of its centre of mass, and be uniformly accelerated with the forces from the ground and upwards.
If the material in the segments is viscoelastic, the amplitude of the wave will be attenuated as it passes through the tissues. The skeletal system have shown to be in some degree viscoelastic in its behaviour. Consequently, there is a damping of impact forces by the musculoskeletal system.
Skeletal alignment refers to the alignment between two adjacent bone segments, that will result in movement of the joint at the loading following impact. This makes some of the kinetic energy of the system to transform into angular motion. The forces at the joints will then decrease from the distal to the more proximal segments.
The movement of the joints due to the skeletal alignment is dependent on the rotational stiffness at each joint, which is regulated by the muscular activation.
Changing the skeletal alignment has been shown to be an effective means of reducing impact forces in humans (Nigg *et al.* 1995).

Objectives

The aim of this thesis was to study the mechanical stress subjected to different levels of the limb at hoof impact. The study of peak decelerations at the hoof wall and the transmission of peak decelerations through the distal bone segments, together with measurements of the ground reaction forces and joint kinematics, were used to describe the characteristics of the mechanical stress subjected to the distal limb.

Hypothesis

The general hypothesis for the thesis was that there is a relationship between the movement pattern of the distal bone segments and the hoof deceleration following first contact. This relationship is supposed to be influenced by the hoof and ground surface interface, and by the speed.

It was also hypothesised that impact related transients transmit through the distal segments of the limb.

Development of the first protocol of methods for measurements

The initial intention of the studies included in the present thesis was to implement tri-axial accelerometers for the measurement of impact related acceleration at different levels of the fore- and hind limb, together with measurements of ground reaction forces and joint kinematics.

As one aim of the study was to measure the transmission of impact related shock waves through the distal bone segments, it was early decided to use accelerometers attached to the distal bone segments. Different methods have been used to attach accelerometers to the skin (Wosk and Voloshin, 1981, Kim *et al.* 1993), however a fixation to the skeleton gives more correct indication of bone accelerations (Light *et al.* 1980, Lafortune *et al.* 1995).

Different designs of attachment for both bone and hoof mounted accelerometers were tried before the final configuration (Paper I). Different types of accelerometers were also considered, in order to find an accelerometer as light as possible and appropriate to attach to bone.

Roepstorff and Drevemo (1993) presented a force measuring shoe that had the capacity of measuring forces in three axes. Two advantages with a force measuring shoe compared to a force plate that was planned to be utilized in the study, is the possibility to use it for overground measurements on different surfaces and at different speeds, and the possibility to use it on the treadmill (Roepstorff, 1997, Roepstorff *et al.* 1994). The first study was conducted on a coir mat treadmill (Fredricson *et al.* 1983), which had the advantage of allowing some movement of the hooves into the treadmill surface. Unfortunately, this setup was not deemed feasible due to technical difficulties.

During this time, Willemen *et al.* (1999) had carried out their study and very generously gave us a bone screw of the same type as they had used. The protocol was further revised to use a force plate for the recordings of ground reaction forces. The hoof and joint kinematics were decided to be captured by the high-speed infrared light emitting cameras that by that time were new at the laboratory.

Materials and methods

Summary of materials and methods in papers I-IV

The horses

All measurements in the present thesis were done with Swedish registered Standardbred trotters. The horses in paper I-III were former trotters that had ended their careers, and were owned by the Department of Large Animal Medicine and Surgery, SLU (paper I) and the Department of Obstetrics and Gynaecology, SLU (paper II and III). The horses in paper IV were all in training and belonged to Travskolan Wången.
All horses were clinically sound and showed no signs of lameness.

Accelerometry

Hoof mounted accelerometers

Tri-axial piezoelectric accelerometers (Brüel & Kjaer type 4504)[1] were used for the recording of velocity changes at the hoof wall. They were mounted on a triangular aluminium plate. The plate was attached to the lateral hoof wall with a two-component glue (Equilox)[2], after the hoof had been cleaned with acetone. The accelerometer was visually oriented with one axis perpendicular to the ground, and one axis parallel to the ground surface at the standing position. The weight of the accelerometer was 14-gram and together with the aluminium plate and a sufficient amount of glue the total weight was 80-95g. The amount of glue varied somewhat for each attachment, partly because of the shape of the individual hooves and the slope of the respective lateral hoof walls. There was no direct contact between the hoof wall and the aluminium plate.
No resonance frequency of the system was experimentally detected below 2400 Hz.
In the first study, one accelerometer was used on the left forelimb. In paper II-IV, accelerometers were mounted on both right fore- and hind hooves.

Bone mounted accelerometers

The same type of accelerometer was used for the measurement of velocity changes on the proximal phalanx and the third metacarpal bone in the first study (see Paper I for details). Two hollow titanium bone screws, 8 mm in diameter, were used for the attachment of the accelerometers to the respective bone segment. These custom-made bone screws were modified from the bone screws used by Willemen *et al.* (1999).
The screws were fixed under general anaesthesia with the horse in right lateral recumbency. The screws were positioned on the lateral side in the middle of each bone segment and perpendicular to the bone length axes.

Each accelerometer was mounted on a hollow connecting-piece of aluminium, fixed on the titanium bone screws between two titanium nuts. One axis of the accelerometer was visually adjusted parallel to the length axes of the bone segment. Due to technical failure, data from the accelerometer on the proximal phalanx were not used in the study.
The total weight of the titanium screw, the two nuts and the aluminium connecting-piece was 27 g. The natural frequency of the attachment device was experimentally determined to be above 1600 Hz when mounted in bone.

Force plate recordings

In paper I-III the ground reaction forces were recorded using a strain gage based force plate (Bertec®, type 6090-15), mounted in a concrete base, half way along a 40 meter long indoor track. The natural frequency of the force plate was experimentally determined to be 175 Hz. The original aluminium top plate of the force platform was covered by a plate made of wood (paper I) or an additional 3 mm thick aluminium plate (paper II and III).
In paper I, a thin layer of sand (0.5-1 cm) was spread out over the covering plate and in paper II a 1 cm layer of sand (particle size 0-5 mm) covered the aluminium plate. The sand was chosen to allow some movement of the hoof at landing (Barrey *et al.* 1991).
In paper III, two different surfaces were used to cover the force plate, in order to modify the hoof and ground interaction properties. One surface consisted of sandpaper, which was glued to the aluminium plate and the other surface consisted of a 1 cm thick layer of sand (particle size 0-5 mm), spread out over a sheet of nonwoven, covering the sandpaper.

High speed videography

In paper I-III movement data were captured using a 240 Hz video system (ProReflex®, Qualisys), with 3 infrared emitting cameras positioned to allow reconstruction of the 3D position of each marker. Spherical reflective markers (11 mm in diameter) were glued on the three corners of the triangular aluminium plate also used for the accelerometer. For the marker positions on the limbs see paper I and III, respectively.
The speeds were calculated from a marker attached to the lateral side of the chest (paper I-III).

Data acquisition

In paper I-III force plate and accelerometer data were sampled at 4.8 kHz using a 16-bit A/D-converter, together with a synchronizing signal from the 240 Hz camera system. Data were captured during 3 s, in order to cover one forelimb stance phase (paper I) or one fore- and hind limb stance phase (paper II and III) over the force plate.
In paper IV, accelerometer data and the signal from a revolution counter mounted on a wheel of the jogcart to calculate speed, were sampled at 10 kHz using a 16 bit A/D-converter during 10 s, which covered at least 10 strides.

Acquisition and processing of data were carried out in the LabVIEW[5] software package.

Experimental design and protocol

Paper I

Two unshod horses, bodyweight 500 and 460 kg, respectively, were used in this study. The surgery and analgesia are described in paper I.
The horses were trotted by hand over the force plate at their own comfortable speed. At least 5 successful left front hoof ground contacts were recorded in each horse.

Paper II

Seven unshod horses were used in this study. The horses were trotted by hand over the force plate. The speed was varied within all horses (the minimum variation in speed within a horse was within a range of 0.6 ms^{-1} and the maximum variation was within a range of 1.6 ms^{-1}). Within the group the speed range was 3-5.3 ms^{-1}. For each horse 10-16 successful right fore- and hind hoof ground contacts, respectively, were recorded in total.

Paper III

Seven unshod horses participated in this study. The horses were trotted by hand over the force plate, first covered by sandpaper and thereafter by sand. A total of 10-16 successful right fore- and hind hoof ground contacts were measured on each surface at different speeds. The speed range of the lowest speeds in individual horses was 0.6 ms^{-1} and the maximum speed range was 1.6 ms^{-1}. The speed range in the whole group of horses was 3-5.7 ms^{-1}.

Paper IV

Eight horses, all shod with standard 6 mm iron shoes were used. The horses were equipped with a regular harness and a head check and trotted at different speeds in front of a jogcart. The mean weight of the horses was 462 kg (SD 43, range 416-546 kg).
The data acquisition was carried out along one straight of a 1 km oval harness training racetrack. The track surface was stone dust and the turns were banked. The recordings were carried out during two following days with about the same dry weather. The horses were earlier accustomed to trot at slow speeds in right turned laps (clockwise) and at high speeds in left turned laps (counter-clockwise), which was also the case in the experiment.
Before the experiment, the horses were warmed up (3 laps right turn at slow speed, approximately 4 ms^{-1}). the The driver was instructed to keep the requested constant speed with a minimum of interference with the horse. Each recording was initialized telemetrically from a car and stopped automatically after 10 s.

The first recording was performed at 10.1 ms^{-1} (SD 0.9) with 4 standard calks (height 9 mm, sides 10x10 mm) mounted under the shoes, after which the calks were dismounted. The following runs were at 4.7 ms^{-1} (SD 0.6), 5.7 ms^{-1} (SD 0.6), 9.8 ms^{-1} (SD 0.8) and at the maximum speed of the horses 12.7 ms^{-1} (SD 0.7).

Signal analysis

The force plate data were filtered back and forth by applying a digital third order low-pass Butterworth with the cut-off frequency 85 Hz (paper I-III) and the accelerometer data were filtered with a second order digital low-pass Butterworth filter at a cut-off frequency of 1200 Hz. In paper II-IV an additional 2nd order digital low-pass Butterworth filter was used on data to get support files used for a consistant selection of variables. In paper I and IV the cut-off frequency of this filter was set to 200 Hz and in paper III to 400 Hz.
In order to get an entire view of the acquired signals, the frequency content of the accelerometer data was determined from an integrated power spectrum analysis (for details see paper III and IV). Before this analysis, the accelerometer data were first filtered at a cut-off level just below the respective Nyqvist frequency. The integrated power spectrum of each study was used to estimate how much of the signal power was below 1200 Hz (paper I-IV). In paper III and IV, it was also to compare the characteristics of data. In these studies, it was assessed how much of the signal power was below certain frequencies, which was of interest in the comparisons.

Variables

The characteristic pattern of decelerations at the hoof, in relation to ground reaction forces was first recognized in paper I (see results paper I in this thesis).
In paper II-IV somewhat different names were used for the variables of the accelerometer signals, while only hoof mounted accelerometers were used.
Amplitudes and times for a number of typical and consistent maxima and minima were defined in accelerometer data, and in paper II and III also in force plate data.
First contact was normally defined from accelerometer data as the beginning of the first vertical deceleration peak (Hjertén and Drevemo, 1994). In some cases when this peak was not evident, first contact was defined as the onset of the vertical GRF (Schamhardt and Merkens, 1994).
z_1 and z_2 were defined as the first and second distinct vertical deceleration peaks at hoof impact. The minimum horizontal deceleration following z_1 but in the horizontal deceleration was denoted y_{min}. y_{max} was the maximum horizontal deceleration, found in the peak retardation complex following y_{min}.
The end of the hoof braking was identified from the horizontal deceleration curve as the time where the hoof was supposed to have stopped moving in the horizontal plane.
In the vertical force signal, there were sometimes well defined local maxima on sandpaper. As these maxima were shown only occasionally, maximum loading rates (DF_{z1} and DF_{z2}) were used instead to describe the vertical force development (Hennig and Lafortune, 1991).

The maximum cranio-caudal GRF was denoted F_{ymax} and the following minimum was F_{ymin}. The maximum loading rate to F_{ymax} is DF_{ymax}. The time for first contact was set to zero.

Statistics

For information on the statistical methods used in the papers, see the respective "materials and methods" and "results". See also the discussion about why the methods were chosen in "methodological considerations" below.

Methodological considerations

In paper I-III three, separate measurement methods are used synchronously. In paper I-III the high-speed videography was used to determine the speed of the horses over the force plate. In paper I, it was also used to assess the hoof angle at impact. Both in paper II and III many of the recordings on the sand surface suffered from missing data, due to dust, which disturbed the infrared reflection from the hoof markers.

Considerations on the attachment of accelerometers

For the attachment of transducers to bone with intraosseous pins, Rostedt *et al.* (1995) certified that it is of great importance that the resonance frequency of the system is considerably higher than the frequencies of interest. In paper I, the resonance frequency was experimentally tested by mounting the accelerometer system to the distal limb of a horse cadaver. The resonance frequency was found to be higher than 1700 Hz.

When considering accelerometer signals, it is important to keep in mind that the signal may not only reflect the velocity changes of the bone, but also other influencing factors (Lafortune, 1991). For example, skeletal alignment is important to consider. The loading of a segment is determined by the acting forces and moments. The accelerations are a result of the input force on the most distal segment and the geometrical alignment of the chain of segments (Nigg *et al.* 1995). When measuring acceleration of a specific location in a segment, the signal is composed of a summation of the accelerations due to impact, the angular motion of the segment and gravity (Lafortune, 1991). This could be interpreted as rotational, translational and gravitational components (Nigg *et al.* 1995). All three components of acceleration must be measured to quantify the full effects of impact on bone segments during locomotion (Lafortune, 1991). The effect of the angular motion of a segment depends on the distance between the accelerometer and the centre of rotation for the respective joint (Lafortune & Hennig, 1991). Therefore, the accelerometer was positioned in the middle of the bone segments in order to achieve a minimal influence from the rotation of the joints.

Considerations on filtering procedures

Force plate data were filtered back and forth applying the same procedure in paper I-III, using a digital third order low-pass Butterworth filter. The cut-off frequency was 85 Hz. The natural frequency of the force plate was tested to be 175 Hz. The order and cut-off frequency were decided to be appropriate after visual examination.

In paper I, FFT analyses of accelerometer data showed that approximately 99 % of the signal power was in below 1200 Hz. Therefore, this was decided to be an appropriate cut off frequency. The second order filter was chosen after visual examination that revealed minimal changes in the temporal information of the time series.

In paper II, the same filtering procedure was used as in paper I. A second order digital low-pass Butterworth filter with a cut-off frequency of 200 Hz, was used as a support for a consistant selection of variables.

A power spectrum analysis of the accelerometer data revealed that after filtering just below the Nyqvist frequency (2395 Hz order 2), approximately 95% of the signal power was below 1200 Hz.

In paper III, it was decided to use the same filter as in paper I and II to make comparisons of data easier. A 2nd order digital low-pass Butterworth filter at a cut-off frequency of 400 Hz, was used to support the selection of variables.

There question arose if the second order filter at 1200 Hz used to determin the variables, was steep enough not to reduce the higher deceleration peaks when the characteristics of the two surfaces were compared. This apprehension was confirmed when a 1200 Hz order 10 was tested. However, this problem was judged not to influence on the conclusions.

In order to get an entire view of the acquired signals, the accelerometer data were filtered back and forth using a 10^{th} order low-pass Butterworth filter, with the cut-off frequency set to 2399 Hz (just below the Nyqvist frequency) (Burn *et al.* 1997). The frequency content was then determined from a power spectrum analysis. The power spectrum was integrated to estimate the frequencies at 50% and 90 % of the signal power (Lanovaz *et al.* 1998).

In paper IV only accelerometer data was acquired. Also in this study, the second order Butterworth at a cut-off frequency of 1200 Hz was apllied. The integrated power spectrum showed that approximately 60-80 % of the signal power was below 1200Hz. A second order digital low-pass Butterworth filter at a cut-off frequency of 200 Hz, was used to support the selection of variables.

Statistics

Considering the data with the wish to find out whether there are differences between different set ups or not, two main problems first had to be solved. It was about the variation between the individual horses and about the variation within each individual horse.

Before a decision is taken about the method to use to compare data from different set ups, it is important to consider the intra- and interindividual variation.

When treating data after a paired design as dependent samples, it is not necessary to take the variation between individuals into account.

The intraindividual variation is more difficult to handle. For example in paper IV, a number of successive strides were recorded. Due to variations in how the horse places the hoof on the ground (Fredricsson *et al. 1972*) and possible inconsistencies in the ground surface a certain variation is introduced. In paper I-III only one step per run was recorded by the force plate, which makes the results sensitive to differences in speed.

Therefore, in paper II and III the data of each step was handle separately in order to reduce the intraindividual variation as much as possible. The intraindividual variation was not accounted for in paper IV. Instead the means of the range of steps were representing the individual at each speed.

When the aim is to compare measurements from the same individual but at different set ups, the use of paired data is a sufficient method (Altman, 1991). Paired data are used to study the average differences between the observations for each individual and for the variability of these differences. The strength of this design is that the interindividual variability is removed. The differences are used to calculate a P value for the comparison of means. If the differences have a normal distribution the one sample *t* test can be used, and in other cases a Wilcoxon test can be applied. In the present study the two methods lead two similar results, but with the difference that the P value showed a tendency to be larger when using the Wilcoxon test. This tendency, however, is a general tendency for small samples (Altman, 1991).

Results

The transmission of peak decelerations through the distal segments of the forelimb (Paper I)

The characteristic pattern with the variable names marked out of the accelerations at the hoof, the accelerations at the metacarpus and the ground reaction forces at the beginning of one stance period of one horse are shown in paper I, Figure 1.

Following first contact, there was a rapid increase in the longitudinal retardation of the hoof, which appeared as a single peak (H_zG_1). The first peak indicating a horizontal velocity change at the hoof (H_yG_1), showed a large variation in amplitude and timing, though the change of rate towards the peak always began at FC. The first longitudinal velocity change of metacarpus ($McIII_zG_1$) appeared about 1-3 ms after H_zG_1 and H_yG_1.

About 4 ms after H_zG_1, the next longitudinal hoof retardation peak (H_zG_2) appeared at the time of the second distinct increase in the horizontal hoof braking (H_yG_2) and metacarpal retardation ($McIII_zG_2$). From the frontal and lateral hoof angles, full support occurred toward the end of the second complex of the hoof accelerometer (Figure 2). Timing of events generally seemed to be shorter in horse A than in horse B.

The mean values of magnitudes and time of the deceleration variables for the two horses are presented in Table 1. Each mean value has been calculated from the parameter value obtained of the individual curves.

Within individual variation was calculated for each subject by determining the standard deviations of all parameter values across the six repetitive trials.

Figure 2 shows the horse B vertical (F_z) and horizontal (F_y) forces at the hoof, together with the longitudinal acceleration at the hoof.

The curves in Figure 2 are normalised to the difference in time between FC and the $F_y max_1$ divided by the difference in time between FC and $F_y min_1$. Only the initial part of the stance phase is visualised.

In horse A, the $F_z max_1$ and the $F_y min_1$ of the vertical and horizontal force traces coincided with the oscillations of the hoof and metacarpus, before the hoof was in full contact with the ground. In horse B, $F_y min_1$ coincided with full support.

In horse A, $F_z max_2$ and $F_y min_2$ coincided with the end of horizontal hoof braking, at the same time there was a local maximum in the slope of the lateral metacarpal angle. These events are followed by a local minimum of the F_z trace ($F_z min_1$), a local maximum of F_y trace ($F_y max_1$) and a local minimum of the lateral metacarpal angle. In horse B the $F_y min_2$ coincided with the end of the horizontal hoof braking, when the slope of F_z and the metacarpal angle were slowly increasing.

Differences in the hoof braking between fore- and hind limbs (Paper II)

Examples of the characteristic patterns of the vertical and cranio-caudal decelerations, and the vertical and cranio-caudal GRFs during the hoof braking period are shown in paper II Figure 1. Both the vertical (F_z) and horizontal (F_y) load were characterized by slow onsets, followed by an increased force. The F_z curve showed inflections following both DF_{z1} and DF_{z2}. Compared to F_z, F_{ymax} was located between DF_{z2} and the following inflection. The presence of a corresponding inflection in the F_y curve varied.

At FC, in the horizontal deceleration curves, an initial complex of more or less prominent deceleration peaks was followed by the distinct local minimum y_{min} that indicated a short period of acceleration of the sliding hoof. The time of y_{min} coincided with DF_{z1}. The z_2 peak deceleration complex appeared at a time between the beginning of the hoof acceleration and y_{min}. The y_{max} peak deceleration complex followed immediately after y_{min}, which indicated a fast braking of the hoof. The y_{max}-complex coincided with the inflection in the F_z curve, and if present also with the inflection in the F_y curve. The end points of the TCP were located between F_{ymax} and F_{ymin}.

The one-way ANOVA carried out for each variable and limb showed significant inter-individual differences ($P<0.001$) for all variables studied. The means and standard deviations of the differences between the fore- and hind limb variables are presented in Tables 1 and 2. The standard deviations revealed large intra-individual variations.

No significant differences in accelerometer data between fore- and hind limbs were observed in the group of horses. In the force plate data significant differences were observed for $DF_{z2,}$ F_{ymax} and the time for F_{ymin} (Figure 2 and Table 2), which was confirmed by the Wilcoxon matched pair- and the single sample t-test ($P<0.001$).

The hoof braking on two different surfaces (Paper III)

Significantly higher frequencies in the vertical and horizontal decelerations were seen on the sandpaper surface compared to sand (Figure 1). On sandpaper in 5 out of the 7 horses, a dominating part of the signal density was concentrated to the lower frequencies in the forelimb, when compared to the hind limb. However, the accelerometer data did not show any significant differences between fore- and hind limbs.

No correlations were found between speed and the different variables.

The characteristic patterns of both the vertical and cranio-caudal GRFs and the decelerations on sand were more distinct on sandpaper in all horses during the hoof braking period (Figure 2).

Significant differences were seen in the deceleration patterns between the two surfaces and means and standard deviations are summarized in Tables 1 and 2. A shorter time between first contact and z_1, and a higher y_{max} were observed on sandpaper compared to sand. In addition, z_1 in the forelimb was significantly higher on sandpaper. An initial complex of peaks was seen in the horizontal deceleration signal, which was most prominent on sandpaper.

The vertical load was characterised by a slow onset and the following fast increase started earlier on sandpaper compared to the sand surface. On sand, the F_z curve showed inflections following both DF_{z1} and DF_{z2}, while the corresponding events were seen as distinct and consistent local maxima and minima on sandpaper in a majority of the recordings.

Significant differences were observed for times and magnitudes of F_{z2} between surfaces.

DF_{z1} was higher in the forelimb (and showed a tendency to be higher also in the hind limb) and the maximum horizontal force appeared significantly earlier and showed higher magnitude on sandpaper compared to sand. DF_{ymax} was higher on sandpaper compared to sand in both fore- and hind limbs.

F_{ymax} was located between DF_{z2} and the following inflection on sand but was seen earlier in the recordings on sandpaper. F_{ymin} appeared after the inflection followed after DF_{z2} on both surfaces.

In the kinematic data, in a majority of the impacts on sand were missing values from first contact and through the initial part of the hoof braking, probably due to dust covering the markers. There was not enough data for a complete analysis to be considered. However, the complete recordings that was captured revealed a pattern of the fetlock joint angle in accordance with the angle-time diagrams presented by Back *et al.* (1995a, 1995b) and Johnston *et al.* (1995, 1996), with a short period of slow movement of the fetlock joint angle before the rapid extension of the joint at a time that seemed to be in accordance with the rapid increase in the vertical load.

The hoof impact at low and high speed on a harness racetrack (Paper IV)

The deceleration patterns of representative steps are presented in Figure 1. The vertical deceleration curves were most characteristic at higher speeds filtered at 1200 Hz. At low speeds, their typical appearance became more consistent after filtering at 200 Hz.
In the horizontal deceleration curves, the minimum, y_{min} was the most pronounced feature, followed by an often steep increase to the extended y_{max}.
In addition, more or less prominent series of deceleration peaks occurred between FC and y_{min} and occasionally, the maximum value of the horizontal deceleration was found here.
The ranges of the means (Figures 2, 3, 4, 5) indicate certain significant differences between horses, and the overall relatively large standard deviations demonstrated considerable differences between individuals. The ranges of standard deviations were quite similar within variables for the different speeds. Instead of reporting all standard deviations, the variances were surveyed as the ranges of the coefficients of variation.
The pair wise comparisons of the respective means of the 4 variables representing the vertical and horizontal peak deceleration patterns (Figure 2), showed that the magnitudes increased significantly between the lowest and highest speeds.
The coefficients of variation for z_1 for all speeds varied within a range of 32-91 % (Figure 2). For the second vertical peak deceleration, z_2, the coefficients of variation were spread within the range of 33-107 % for all speeds. The corresponding data for y_{min} and y_{max} were 12-98 % and 15-88 %, respectively.
The time variables were generally expressed as percentages of the stance time (Figure 3), and for z_2 and y_{max} also in milliseconds (Figure 5). The stance time became significantly shorter with every step of increased speed. In the forelimb, the time of z_1 as percentage of the stance time was significantly longer at high speeds. The coefficients of variation for z_1 were in the range of 15-81 %. The times of z_2 as percentages of the stance time were significantly longer at high speeds for both fore- and hind limbs and the range of coefficients of variation were 13-40 %. The times for z_2, expressed in milliseconds, were significantly longer in the lowest compared to the highest speeds.

The times for y_{min} expressed as percentage of the stance time were significantly longer with increased speed in the forelimb, but not in the hind limb. The corresponding coefficients of variation were 9-47%.
The time for y_{max} did not change significantly with speed and the coefficients of variation varied between
6-78 %.
The end of the hoof braking expressed in ms did not show significant differences (Figure 4), and the range coefficients of variation was 6-54 %. When expressing the hoof braking period as percentage of the stance time, there was a significant increase with speed, with the coefficients of variation ranging between 5-56 %.
There were no significant differences between 10.1 ms^{-1} with calks and 9.8 ms^{-1} without calks for any variable.

The power spectrum analyses of the data filtered at 4999 Hz showed little variation between- and within individuals in the distribution of frequency content. The frequency levels for 50, 80 and 90 % of the signal power did not show any systematic differences in the individual horses at the different speeds.

Discussion

General considerations

The papers included in this thesis are all part of the work to fulfil the aim, to investigate the deceleration pattern of the hoof following the first contact with the ground at the beginning of the stance phase. The work on the papers has been carried out during several years. During the work new perspectives have arisen leading to some alterations in definitions and terminology.

The general pattern of the hoof deceleration are described in papers I-IV, based on accelerometer recordings in the directions of the vertical and of the hoof (In paper I longitudinal and cranio-caudal horizontal axes). Recordings in the latero-medial horizontal axis were not considered in the study as this direction was assumed to contribute to the resultant deceleration of the hoof only to a minor extent.

In paper I, a typical deceleration pattern was apparent even if the speed was low and the surface allowed some initial movements due to the construction of the treadmill belt.

In paper II the recordings were performed on a surface chosen to permit a relatively free movement of the hoof in the horizontal plane at impact (Barrey *et al.* 1991). The general deceleration pattern did not differ considerably from what was seen in paper I, however obvious differences were observed occasionally in individual horses. Certain differences between fore- and hind limbs were also seen in individual horses. It was found that some basic features became more distinct when a relatively low cut off frequency in the filter was applied, especially for the first two vertical deceleration peaks (z_1 and z_2)).

In paper III there were significant differences in the signal, when two different surfaces were compared. On the sandpaper surface, with assumed higher friction, the pattern was very distinct but on the sand surface, interpretation of the signal peak pattern required more consideration.

In paper IV, the characteristics of the signal were similar to the sand surface at speeds comparable to the speeds in paper II and III. At higher speeds a distinct and consistent pattern was apparent in all horses.

Details of the basic hoof deceleration pattern

First contact and the initial deceleration

First contact was defined from the vertical oriented accelerometer signal (Hjertén & Drevemo, 1994; Merkens &Schamhardt, 1994). The start of the hoof impact was characterised by a distinct vertical and horizontal peak decelerations. This was most prominent in the vertical direction but showed large intraindividual variations in amplitude (paper II-IV). The variation may reflect differences both in the hoof position at the landing between strides and differences in the ground surface properties (Fredricson *et al.*, 1972; Barrey *et al.*, 1991; Benoit *et al.*, 1993; Willemen *et al.*, 1999).
As sound Standardbred trotters in general prefer to land with heels or the rear lateral lateral part of the hoof first (Fredricson and Drevemo, 1972; Dalin *et al.* 1973; Johnston *et al.*, 1996, 1997), it could be assumed that this was the case also in the present studies. This may explain that the first vertical peak deceleration (z_1) was by far the most prominent (paper IV), as the accelerometer if landing heels first was in nearly alignment with the longitudinal direction of the limb. The first peak deceleration was seldom recognizable in the cranio-caudal horizontal deceleration.

The initial vertical peak deceleration was followed by a second peak complex, in which a consistent maximum (z_2) was identified. At about the same time a major acceleration peak was observed in the horizontal signal, in most cases followed by a horizontal maximum peak deceleration complex (paper II-III). This reflects the horizontal hoof braking which coincides in time with a steep increase of both the vertical and horizontal force curves.

The second vertical peak deceleration appeared after 5-10 % of the stance phase. This corresponds to 5-10 ms after first contact (paper I and II). Probably the second vertical peak retardation complex rather reflects the hoof translation towards the ground and the flexion and extension in the coffin and fetlock joints, respectively (Back *et al.*, 1995a, 1995b, Johnston *et al.* 1995, 1996) and can be denoted the concussion period. This suggestion is further supported by the maximum horizontal deceleration appearing in time close to the time of the maximum loading in both the vertical and horizontal forces as well as the angle-time diagrams by Back *et al.* (1995a, 1995b)

After first contact, an initial flexion appears in the coffin joint until full support (Hjertén & Drevemo, 1994; Back *et al.* 1995a, 1995b). The horizontal acceleration was probably caused by a combined of this flexion and a restricted forward translation of the hoof at the initial contact with the ground (paper I).

Force measurements

The force plate measurements presented in paper I-III showed a very low initial loading during the first milliseconds after first contact. This has earlier been observed and suggested to be due to the low effective mass acting at the very initial part of the stance phase, partly due to the flexion of the hoof segment in the coffin joint and a translating movement of the low mass (Hjertén & Drevemo, 1994). Based on the synchronised recordings of force plate and accelerometer data, it would be possible to estimate the effective mass using Newtons 2nd law. A large initial peak deceleration then implied a low effective mass. It must, however, be taken into account that the deceleration amplitudes in the present studies are smoothened by the applied filtering procedure. Nevertheless, the results confirm that only a minor part of the distal segments of the equine limb can possibly be loaded in the initial part of the stance phase (Hjertén & Drevemo, 1993, 1994; Hjertén *et al.* 1994).

In addition, Riemersma *et al.* (1987a, 1987b) showed that the load and strain of the digital flexor tendons and the suspensory ligament have a strong correlation to the vertical loading of the limb. Consequently, the extension of the fetlock joint may indicate the application of an increasing vertical load. It could further be deducted from angle-time diagrams captured by high speed cinematography (Back *et al.*, 1995a, 1995b) that little movement occurs in the fetlock joint during the initial 2-3 % of the stance phase. This supports the suggestion that only the very distal parts of the limbs are in fact loaded in the first 2-3 % of the stance phase.

Simultaneously with the successive increased loading of the limb (Hjertén & Drevemo, 1993; Hjertén *et al.* 1994), and the interaction with the ground surface, the horizontal loading increases and the hoof will be forced into horizontal braking. The appearance of the second vertical peak deceleration complex and the maximum horizontal deceleration peaks are both influenced by the surface characteristics. In paper II and III it was evident that the time for the maximum horizontal deceleration was close related to the time of the loading rate maxima in both the vertical and horizontal force components.

The distal limb as a shock absorber

Impact related transients seems to be effectively attenuated in the distal limb as transients measured on the hoof wall were largely diminished in the bone proximal to the fetlock joint (paper I). It was therefore suggested that the initial vertical peak deceleration may be negligible proximal to the fetlock joint.

It should however be kept in mind that surfaces with different frictional and damping properties may change this conclusion. The properties of the ground surface and speed are important to the impact loading pattern of the limb. It is shown that higher friction between hoof and ground (paper III) results in higher loading rates and peak amplitudes of the maximum horizontal deceleration, which means that higher mechanical stress may be transmitted to limb structures also proximal to the fetlock joint. In addition, higher speed results in higher loading rates and maximum horizontal peak decelerations, as well as higher vertical peak decelerations (paper IV).

The horse has a further natural shock absorbing capacity due to the skeletal alignment that results in successive and prolonged disto-proximal loading in the beginning of the stance phase (Hjertén & Drevemo 1994). A ground surface and optimal shoeing with respect to shock absorption improves the possibilities for the horse to utilize these prerequisites. This is reflected by the second and the maximum horizontal deceleration peaks and the loading rates, which are also considered to be indicators of the mechanical stress subjected to the limb.

Conclusion, application and suggested research

Conclusion

According to the results in the thesis, the hoof deceleration can be divided into two parts, due to the successive loading pattern of the limbs.
The first part is characterized by the often large impact peak. This can be assumed negligible to structures proximal to the fetlock joint, while it is to the most extent attenuated within the more distal structures. The magnitude of the peak varies due to the normal variation of hoof orientation at impact. Though, there is an increase in amplitudes at higher speed.
The second part of the hoof deceleration is characterized by the onset of loading of the fetlock joint, and the interaction between the hoof and the ground surface characteristics. The pattern of the second part is determined by the movement pattern of the distal bone segments and the rate of loading of the limb, but also to a large extent related to the properties of the ground. It is shown that a higher friction results in a higher loading rate and higher peak amplitudes of the maximum horizontal peak deceleration, which means a higher mechanical stress transmitted to the structures of the limb most likely also proximal to the fetlock joint.
Also a higher speed gives higher loading rates and maximum horizontal peak decelerations, as well as higher vertical peak decelerations.
The horse has a natural shock absorbing capacity due to the skeletal alignment that normally results in a successive disto-proximal loading scheme and a prolonged time for the loading uptake.
An optimal ground surface and an optimal shoeing with respect to shock absorption allows the distal limb to utilize its anatomical prerequisites. This will moderate the loading rates, of the second vertical peak retardation and the maximum horizontal deceleration peak.

The results presented in the thesis bring new knowledge to the events of hoof impact and the hoof deceleration following impact. The findings also contribute to the the description of the shock absorbing capacity of the equine limb, and the mechanisms behind indicators of mechanical stress to the structures of the distal limb.

Applications

Development of farrier techniques

The results of this thesis give the basis for further development of horse shoes and ground surfaces. The accelerometer technique has proven to be useful for determining important interaction properties between the hoof and the ground in the early stance phase.

For the training horse to avoid overload

The results may be implemented in the field of horses training in order to prevent from impact related injuries in the locomotor apparatus. The findings increase the knowledge about the importance of the interaction between hoof/shoe and ground. It was shown that different properties lead to different characteristics of the mechanical stress subjected to the limb. Sudden changes in shoeing and the qualities of the ground will increase the risk of overload. A better knowledge on the interaction at impact under different circumstances is fundamental for an optimal adaptation.

For rehabilitation after lameness

The detailed knowledge about the mechanical stress of the hoof and ground interaction in the beginning of the stance phase is essential knowledge in the rehabilitation of injuries to the locomotor system. Tissue injuries in carpus and tarsus and to all tissues distal to these joints are directly affected by the interacting properties between the hoof and ground. Together with consideration of the functional anatomy of the extremities, these are important aspects in the choice of farrier technique, of different shoeing, for suitable ground surface and the training program itself.

Suggested future research

The accelerometer technology was shown to be suitable for field studies. This opens possibilities for further research on varying hoof and ground interactions. Further knowledge on the different training situations will open new possibilities to plan for adaptation of the locomotor system in individual horses to fulfil the requirements to meet the mechanical stress of future challenges.

Locomotor disorders as a problem among training and competing racing horses tend to be a classical problem focused on already in the 1960's and 70's and wrote in the introduction of his doctoral thesis "Obviously, many horses are not capable of withstanding the severe stress to which they are subjected during training and racing, with ensuing locomotor lesions often at young age." Maybe we have learned much on how to train horses, but still, 25 years later lameness due to training and competition is a major problem to racing Standardbreds. There still seems to be a great need for a better understand how to prevent injuries.

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