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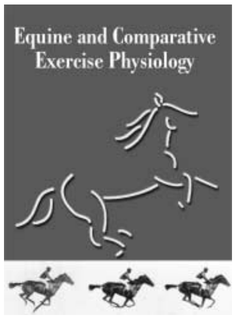

Synthetic shoes attenuate hoof impact in the trotting warmblood horse

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Abstract
Impact is considered the most critical part of the stance phase for the development of chronic articular disorders such as osteoarthritis in the equine distal limb. Modern, synthetic shoeing materials are believed to modify impact and therefore are often used to treat and/or prevent lameness due to chronic joint disorders. Scientific evidence is scarce, however, to prove this. Hoof impact of forelimb was compared quantitatively in a group of horses under three conditions: unshod, classical steel shoes and shod with a synthetic shoe. Twelve sound warmblood horses were trotted by hand on an asphalt track at a mean speed of $3.5\,\mathrm{m\,s^{-1}}$ and measured in a Latin square design (unshod condition, with steel shoes and with polyurethane (PU) shoes) using a triaxial accelerometer that had been fixed to the lateral hoof wall of the left forelimb. The sampling frequency was set at 10 kHz per channel. The maximum amplitude of vertical and horizontal, forward/backward accelerations at hoof impact was lowest when shod using the PU shoeing condition ($P < 0.05$), but the duration of the impact vibrations was lowest when unshod. PU shoes cause more damping, less friction and slower shock absorption at hoof level compared with the other two conditions and thus modify impact. Synthetic, polyurethane shoes may help in reducing peak vibrations. These short-term effects appear to be promising enough to evaluate PU shoes under field conditions in reducing impact on the longer term after substantial wear and tear. Furthermore, the possible role of synthetic materials in repairing critical tissues or even in preventing osteoarthritis in horses warrants further investigation.

Keywords: equine; locomotion; initial ground contact; shoeing; acceleration

Introduction

Lameness is the most important cause of wasting in sport horses. Joint injury and joint diseases are the most common causes of lameness, and together they represent a major part of the caseload for equine clinicians[1]. Immediately after initial ground contact, ground reaction forces rapidly decelerate the hoof and reduce the speed of the distal phalanx of the limb to almost zero[2]. This phase is called the impact phase, which is considered the most critical part of the stance phase for developing injuries of the musculoskeletal system[3]. A repetitive, impulsive loading during running, even within physiologic limits, is considered to play an important role in the aetiology of primary osteoarthritis in man[4,5] and in laboratory animals[6,7], as high-frequency oscillations increase the risk of damage to subchondral bone and other critical joint tissues[8–10]. A similar relationship has been suggested for performance horses[1,11]. The vibrations generated in the horse's limb during each impact are transmitted proximally and attenuated during this process[12–15]. The relative contribution of the different structures in the horse's lower limb to the reduction of these impact vibrations has not yet been clarified. Two basic mechanisms can be distinguished. Structures acting as springs in the leg may substantially reduce impact forces. The more compliant the spring, the better its ability to reduce these forces on impact. Ideally, springs return the energy which is stored in them when they stretch during the subsequent recoil. Energy can also be absorbed by dampers that can turn mechanical energy into heat. Shock reduction by appropriate springs is energetically more effective during locomotion and it has been proved that the leg of a horse behaves like a complex visco-elastic spring with a very good elastic resilience[16].

Factors that influence the initial generation of impact vibrations are the material properties of the two structures that come into contact at hoof strike, i.e. the surface and the hoof itself, or if the horse is shod, the shoe. The characteristics of racetracks with respect to damping properties have already been investigated[17]. Research into modern, special shoeing types, the use that has been advocated to reduce impact vibrations, has been limited so far. It has been based merely on clinical evidence that horses with signs of pain in the distal portion of the forelimb and shod with orthopaedic, damping shoes appear to trot more comfortably[18,19].

Marks *et al.*[18] evaluated clinically a pad elastomer system under the hoof in a group of *c.* 250 horses and estimated that it would reduce concussion by 19%. Vasko and Farr[19] applied accelerometers to hoof, metacarpus, metatarsus, nuchal crest and lumbosacral junction before and after the application of a polymer pad in 34 lame horses with back pain and qualitatively reported a mean reduction in peak energy force of 78% and that 94% of the horses had relief from locomotor abnormalities and had returned to competition. Experiments that objectively assess the effects of synthetic shoeing materials on impact vibrations in sound horses are limited[20] and have never been quantitatively evaluated in a large group of sound horses, so that intra- and inter-individual variability can be objectively compared.

The purpose of our study was to evaluate hoof-impact accelerations in a group of sound horses under different shoeing conditions (unshod, a classical steel shoe and with a polyurethane (PU) shoe) to create a scientific base under modern, equine orthopaedic shoeing materials.

Materials and methods

Horses

Twelve clinically sound Dutch warmblood horses were used in this study. The age range of the horses (ten mares and two geldings) was 5–20 years (mean 9.7 years), the mass ranged from 480 to 632 kg (mean 556 kg) and the height at the withers from 1.60 to 1.71 m (mean 164 cm). Each horse was trained to trot at a constant speed on an indoor asphalt track, while wearing a girth strap to carry the necessary equipment used for data collection.

Shoeing conditions

Measurements were performed with each horse under three shoeing conditions in a Latin square design to prevent influence from the order in which the shoeing conditions were tested. The three shoeing conditions were: unshod, shod with traditional steel shoes and shod with nailed PU shoes (Hippoflex®, CERA Handels GmbH, Gewerbepark-Fürgen 14, D-87674 Ruderatshofen-Immenhofen, Germany, http://www.hippoflex.de; Fig. 1).

Data recording

To minimize the shock-reducing effects of the track, the horses were trotted by hand, on an asphalt track that was considered as a stiff reference surface. A trial was considered successful when the horse trotted along a straight line, with a regular cadence at a mean speed of $3.5\,\mathrm{m\,s^{-1}}$ (± 0.05) in the middle of 10 m of a 20 m-wide asphalt track. For every run, the horse passed the infrared beam of the first pole at the beginning of the recording zone of the track, so that the chronometer was switched on. When passing the second pole at the end of the recording zone, the chronometer was switched off again. The average speed of the horse during every consecutive run ($\mathrm{m\,s^{-1}}$) was calculated by dividing the distance between beginning and end of the recording zone of the track (10 m) by the recorded time (s).

Data from trials that did not meet these criteria were discarded. Results from at least six successful trials of each horse under every shoeing condition were collected.

Accelerations were measured with a triaxial piezoelectric accelerometer (Brüel & Kjaer, type 4326, Skodsborgvej 307, DK-2850 Naerum, Denmark) (size $21 \times 16 \times 9$ mm, mass 10 g), which was fixed to the lateral side of the left front foot (Fig. 2).

According to manufacturer's specifications, the accelerometer records sinusoidal accelerations up to $45\,000\,\mathrm{m\,s^{-2}}$ in the frequency range 0.10–16 kHz[15]. The accelerometer was bolted to a custom-made aluminium mount. After hoof trimming, the lateral side of the left front hoof of each horse was cleaned with ethanol and another aluminium bracket was secured using two-component glue (EquiThane Superfast®, Vettec, Zonnebaan 14, NL-3584 EC Utrecht, The Netherlands). To ensure the consistency of placement, this bracket was fixed at maximum hoof width, the ventral side 7–8 mm proximal to the bottom of the hoof, with one axis perpendicular to the ground. After the horses were shod, the accelerometer with aluminium mount was connected to the hoof. The mass of accelerometer and connecting pieces was 19.9 g, which represents *c.* 2% of the mass of the distal phalanx[2].

The accelerometer was connected to a charge amplifier on the back of the horse through double-shielded cables (type AO 0283/2.5 m, Brüel & Kjaer, type 4326, Skodsborgvej 307, DK-2850 Naerum, Denmark) to prevent cable noise. This amplifier was connected to

FIG. 1 A typical example of a left front hoof shod with a PU shoe: note the bar on the frog

another charge amplifier by a 31 m cable. A person running beside the horse carried this cable. The signal from this charge amplifier was A/D converted and fed to a standard personal computer. The data were collected with custom-made software at a sampling frequency of 10 kHz for each acceleration component. Horses were filmed using a standard home-video camera at the end of the track for retrospective control of data. Irregular trials were identified *post hoc* from the video and eliminated.

Data analysis

From the six successful trials of each horse under every shoeing condition, the second stride was selected. The acceleration–time curve of the impact was analysed using Microsoft® Excel software to calculate different variables. Accelerations during impact (and the unmeasured forces associated with them) were measured in three orthogonal directions: proximo-distal (a_z, corresponding roughly to vertical upward/downward), cranio-caudal (a_y, corresponding roughly to horizontal forward/backward) and mediolateral (a_x, corresponding roughly to transverse inward/outward direction). Accelerations and forces directed upwards, forwards and lateral were denoted positive. The start of the maximum vertical deceleration peak was taken as the beginning of impact. The end of impact was defined as the first point after the peak from where the vertical acceleration amplitudes had decreased to 10% or less of the maximum deceleration peak.

The following variables were obtained from each impact peak[12].

The maximum amplitude in vertical ($a_{z\max}$) and horizontal/forward–backward direction ($a_{y\max}$) represented the maximum deceleration recorded at hoof impact. Acceleration values immediately preceding these peaks were subtracted from the maximum amplitudes to correct for drift. The square root of the sum of the squares (RMS) of the three orthogonal recordings (a_x, a_y, a_z) revealed a_{total}. The maximum value of this absolute deceleration amplitude is denoted as a_{totalmax}.

Impact duration (t_{impact}) was the time between the beginning and the end of impact. Stride duration (t_{stride}) was measured as the interval between the impact starts of two successive strides.

The amplitude distribution of the frequencies was found by Fourier transformation of the signal a_{total}. Values were related to the highest value overall, which was designated as '1', resulting in the mean relative amplitudes of the frequency spectrum on impact under the three shoeing conditions.

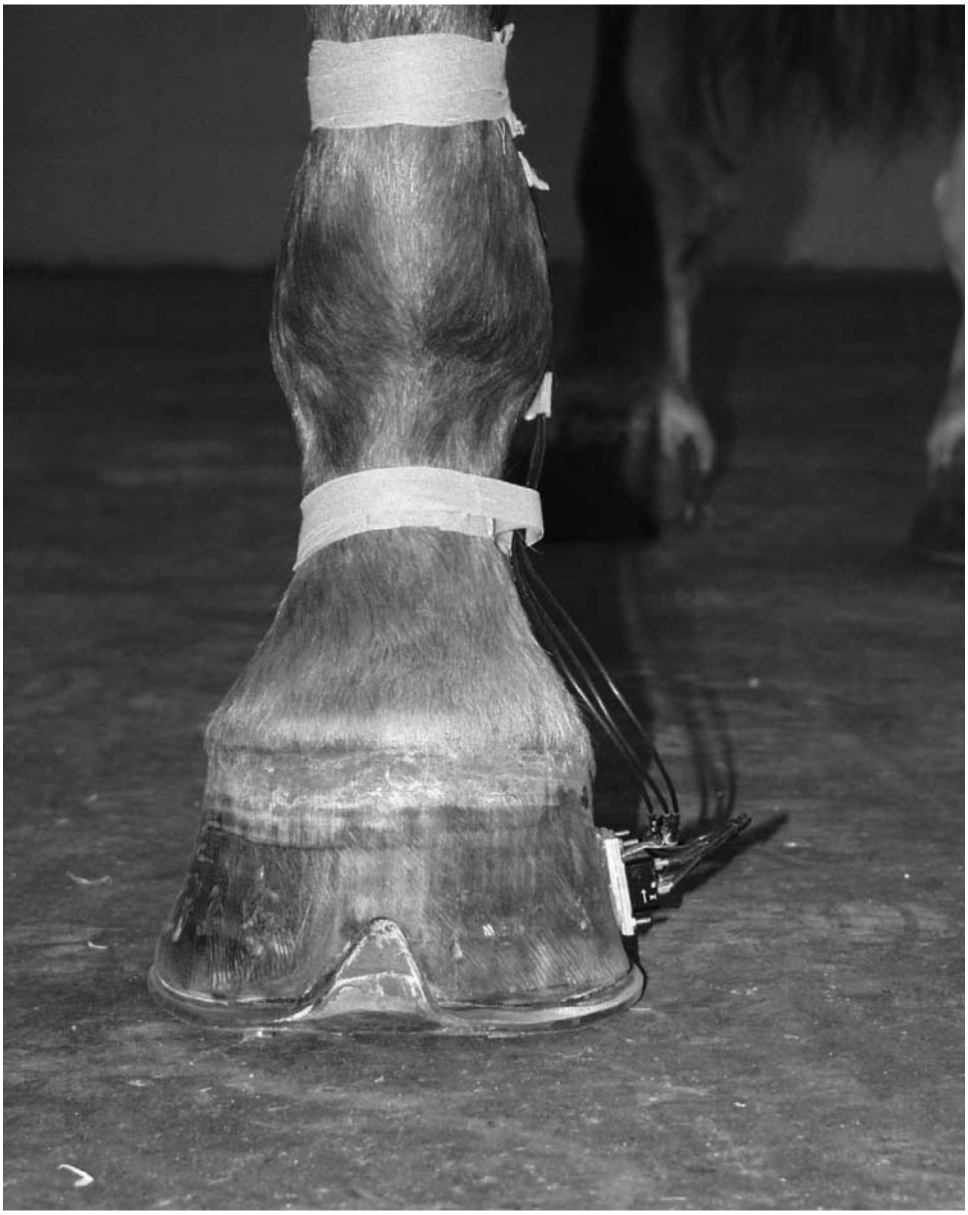


FIG. 2 Accelerations were recorded using a triaxial piezoelectric accelerometer, which was fixed to the lateral side of the left front foot

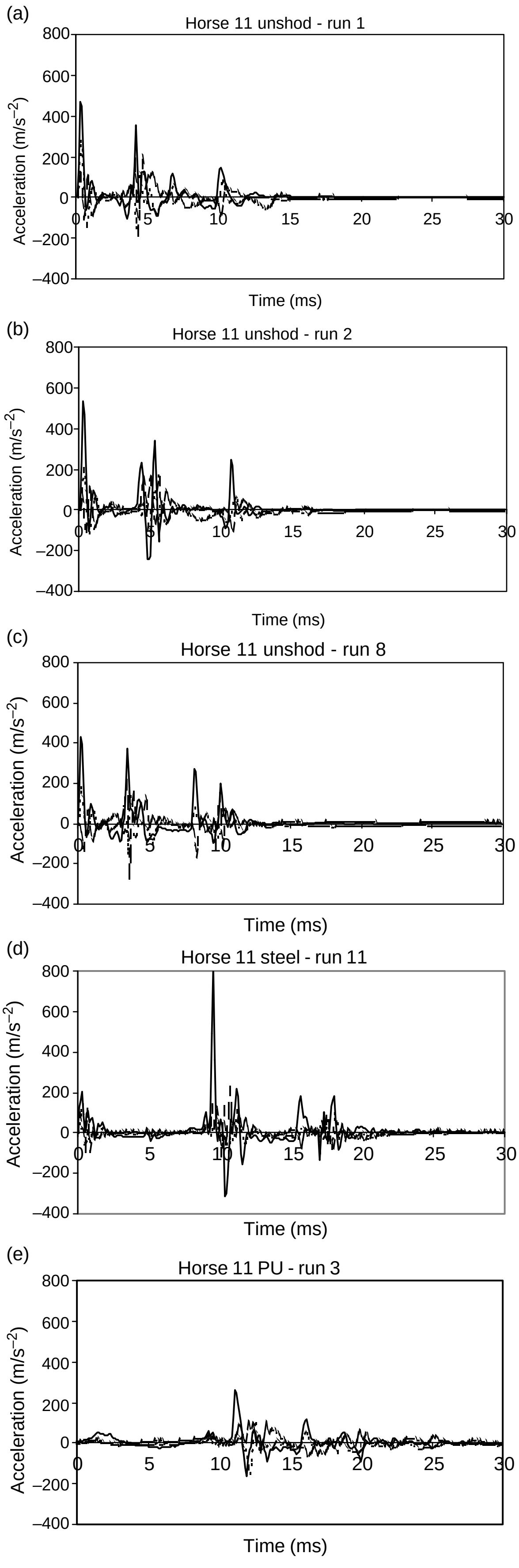
Statistical analysis

The mean maximum absolute accelerations ($\pm$ SD) were calculated by averaging peak measurements of six impacts of each horse under each shoeing condition. Mean impact duration and stride duration were calculated. Differences between the shoeing conditions were statistically evaluated using a repeated measures ANOVA ($P < 0.05$) and a Bonferroni *post hoc* test.

Results

The baseline data confirmed that there was zero change in acceleration with the horse in standing position. The accelerations at the hoof in the initial part of the stance phase showed a characteristic pattern. Following initial ground contact, there was a rapid increase in the vertical and horizontal retardation of the hoof, which appeared as simultaneous peaks ($a_{z\max}$ and $a_{y\max}$). These oscillation patterns in subsequent runs and strides of the same horse were qualitatively evaluated and showed a striking similarity (Fig. 3). Nevertheless, different shoeing conditions could also induce in individual horses a difference in the number of peaks (Fig. 4).

Mean amplitudes of the vertical and horizontal acceleration components of the 12 horses under the three different shoeing conditions are presented in Table 1. The maximum deceleration amplitude was in the vertical direction. In most cases, it was more than twice the amplitude in the forward–backward direction.



The mean maximum deceleration amplitudes ($a_{z\max}$, $a_{y\max}$ and a_{total}) of the PU shoe were significantly lower than those of the steel shoe and unshod condition ($P < 0.05$, Table 2). Differences between steel shod and unshod condition were not significant.

The duration of the impact vibrations (t_{impact}) was significantly shorter in the unshod condition, when compared with the other two shoeing conditions ($P < 0.05$).

The stride duration (t_{stride}) was significantly longer in the steel shod condition ($P < 0.05$).

The mean relative frequency during impact was significantly higher for the steel shoes (0.44) than for the other conditions (0.29 and 0.25, $P < 0.05$; Fig. 5).

Discussion

In this study, accelerations of the equine hoof on impact in horses at trot were recorded with resultant routine values similar to those known in the literature. The variability in maximum amplitude in the three main directions is large, which is in accordance with the findings of Burn *et al.* and Gustås *et al.*[13,21] The intra-individual variability was smaller than the inter-individual variability, while horizontal decelerations (the maximal horizontal force on the hoof owing to friction) show more variability than vertical decelerations. Even though we used only 12 horses and six trials per horse, it was still possible to show a significant difference between the shoeing conditions. Recordings would have been made more repeatable with horses running on a treadmill, but to clearly demonstrate differences between the three shoeing conditions, an asphalt track was needed as a reference track. Previous hoof-acceleration studies have described vertical deceleration amplitudes in the same range as those reported here[13,14,17,20]. In the present study, the mean maximum deceleration amplitude in the vertical direction was $688\,\mathrm{m\,s^{-2}}$ (± 293) for steel shoes and $504\,\mathrm{m\,s^{-2}}$ (± 219) for the unshod condition, though this was not significantly different. See a typical example in Fig. 4. Benoit *et al.*[20] showed a difference between unshod hooves and steel shod hooves at the $P < 0.01$ level, but this was based on only two horses with three to six strides, which does not seem to be an appropriate number of horses for statistics.

Apparently, the vertical mean maximum amplitude for the PU shoe ($343 \pm 251\,\mathrm{m\,s^{-2}}$) was significantly lower than that of the other two conditions, indicating

Fig. 3 Typical similarity in vertical (____, a_z), and horizontal (= forward–backward - - -, a_y) and lateral (. . ., a_x) accelerations during hoof impact of horse no. 11 in three runs (a), (b) and (c) in the unshod condition, in (d) steel shoe and (e) PU shoe condition

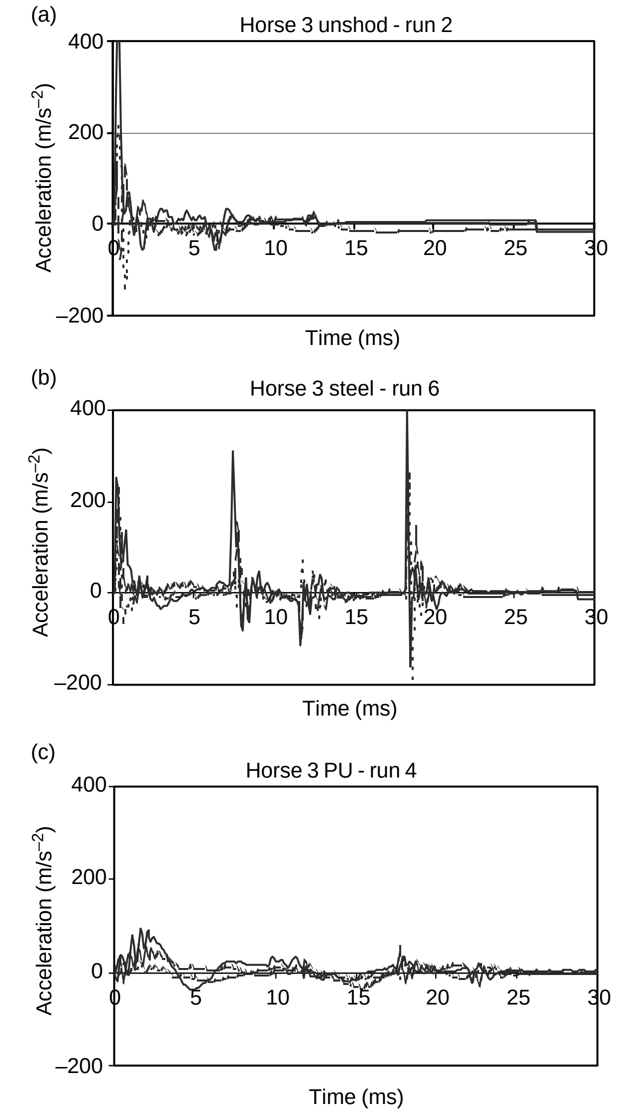


FIG. 4 Typical difference in vertical (____, a_z), horizontal (= forward–backward - - -, a_y) and lateral (. . ., a_x) accelerations during hoof impact of horse no. 3 in (a) unshod (b) steel shoe and (c) PU shoe condition

a significant effect of this synthetic shoeing material. The values for the PU shoe found in this study, however, are still somewhat higher than the $188 \pm 55\,\mathrm{m\,s^{-2}}$ recorded for the shoes with the highest damping in the study of Benoit *et al.*[20] That value was found for the aluminium Equisoft® shoe combined with a PU easy-glue pad. The PU shoes used in the present study were not glued but nailed to the hoof wall. Nails might lead to some loss of damping effect, as the nails will directly pass on the vibrations from the shoe to the hoof wall. On the other hand, the extra bar of the PU shoes might have given some extra frog support and thus contributed to the damping properties of these shoes.

Horse no. 8 showed a remarkably different reaction to the PU shoes, with respect to its response to the different shoeing conditions; the amplitude (in vertical direction only) was larger for the PU shod condition compared with both other conditions (Table 1). Horse no. 7 showed the highest amplitude in the unshod condition. This pattern was not consistent in every trial and could not easily be linked to visual gait abnormalities. The shift found in a number of peaks (Fig. 4), for example, from unshod (one peak) to steel shod (three peaks) might be related to an adaptation in landing technique of horse no. 3 from symmetric, flat to asymmetric, heel–wall–toe[22], thus not exceeding a particular individual maximal acceleration and vibration threshold levels of critical tissues in its limbs[23]. These findings illustrate inter-individual differences in response to shoeing and are observations that should be kept in mind while evaluating the effects of different shoeing techniques for the treatment of individual patients.

The locomotion system is subjected to repeated impulsive loading, as is apparent in vertical and in horizontal ground reaction force traces[24]. Parameters that affect horizontal braking of the hoof may be related to injury of the distal joints[25,26]. The possible factors influencing horizontal loading are forward velocity of the horse, different farrier techniques, structure of the ground surface and shoeing material. The caudal horizontal acceleration peak is required for the hoof to become stationary, and this peak corresponds with maximal hoof friction.

Horizontal acceleration peaks of the hooves have been studied earlier by Back *et al.*[25,27] They found that the horizontal hoof-acceleration peak comes immediately after the vertical hoof-acceleration peak, which is in accordance with the results of our experiment. Only a few data on the maximum horizontal acceleration amplitude during impact are currently available in the literature, and measurements were done under different circumstances. So, it is difficult to compare with the results of the present study. In this study, the horizontal decelerations were significantly smaller in the PU shod condition ($105\,\mathrm{m\,s^{-2}}$), when compared with the steel shod ($206\,\mathrm{m\,s^{-2}}$) and unshod condition ($202\,\mathrm{m\,s^{-2}}$, $P < 0.05$). This means that PU shoes cause less friction. A certain amount of friction is necessary to prevent the hoof from excessive sliding[13]. Both too much friction and sliding may cause joint injuries. Friction depends on the surface characteristics of the ground and the properties of the shoe. Unfortunately, the optimal friction is not known. Probably, the friction of the PU shoes was too low for an asphalt surface, because two horses were seen sliding at the end of the track while slowing down. Probably, applied PU with a lower shoe value will cause more friction (and more damping). However, the material should be strong enough to provide the shoe stability and prevent excessive wear of the shoe. Here, it should be mentioned that testing

Table 1 Maximum accelerations (m s^{-2} mean ± SD) recorded at hoof impact of six strides of the 12 individual horses in (a) the vertical direction (a_z) and (b) the horizontal, forward–backward direction (a_y)

Horse no.	Unshod	Steel shoe	PU shoe
Vertical direction (a_z)			
1	284.9 ± 88.5	518.5 ± 121.5	249.3 ± 38.3
2	451.3 ± 73.7	634.9 ± 134.3	145.9 ± 68.4
3	444.5 ± 92.7	402.5 ± 67.4	81.4 ± 23.2
4	409.5 ± 43.0	645.5 ± 121.6	305.1 ± 92.7
5	632.8 ± 105.8	863.7 ± 413.4	500.7 ± 118.2
6	480.8 ± 72.9	695.8 ± 230.3	509.1 ± 68.2
7	770.0 ± 195.2	513.8 ± 247.1	243.1 ± 102.9
8	899.4 ± 234.6	694.5 ± 132.8	972.0 ± 342.7
9	236.1 ± 43.7	619.5 ± 138.9	222.6 ± 24.1
10	588.5 ± 212.9	688.2 ± 410.9	280.7 ± 53.7
11	493.2 ± 103.8	989.9 ± 295.7	354.8 ± 83.9
12	361.5 ± 45.1	990.1 ± 386.4	254.5 ± 46.2
Mean	504.4 ± 219.1	688.1 ± 292.6	343.3 ± 251.3
Horizontal, forward–backward direction (a_y)			
1	151.9 ± 48.4	170.2 ± 19.8	124.2 ± 37.2
2	163.3 ± 52.3	167.2 ± 26.1	63.5 ± 24.8
3	98.6 ± 62.5	108.0 ± 45.4	43.8 ± 21.3
4	147.0 ± 25.3	229.0 ± 43.4	104.0 ± 49.6
5	314.9 ± 81.4	255.4 ± 138.5	164.3 ± 55.2
6	421.4 ± 122.5	213.9 ± 74.2	209.0 ± 114.3
7	292.2 ± 103.3	214.2 ± 69.1	130.4 ± 60.3
8	292.3 ± 179.1	264.7 ± 92.0	84.2 ± 24.6
9	181.0 ± 18.1	215.0 ± 48.0	78.1 ± 15.5
10	155.8 ± 43.0	191.1 ± 60.9	81.1 ± 41.2
11	117.7 ± 45.9	194.2 ± 50.1	84.6 ± 24.5
12	89.5 ± 20.4	253.3 ± 84.2	93.4 ± 40.0
Mean	202.1 ± 124.3	206.3 ± 77.1	105.0 ± 63.4

immediately after shoeing, as practised in this study, will have influenced friction. In the steel shoes, the nails were still protruding, enhancing friction to a level similar to the unshod condition. The PU shoe had a thin coating that wears off quickly, but may have decreased friction of the freshly shod hooves.

After the impact, the vibration acceleration was progressively damped in 8–15 ms, which is similar to that found in human walking[28], and somewhat shorter compared with the results in other equine experiments[17,20,21,25]. The attenuation of the vibration took markedly longer in both shod conditions (compared to the unshod condition), which may have been caused by the restriction of hoof wall movement due to the nailing[12].

Stride duration was significantly longer in the steel shod condition (701 ± 33 ms) compared with the PU shod and unshod conditions (688 ± 32 and 685 ± 32 ms, respectively); in comparison, Back and Clayton[2] reported a mean value of 670 ms for the treadmill at 4 m s^{-1}. The accelerometer signal was about zero during stance phase, but changed very gradually and inconsistently at the beginning of the swing phase. The longer stride duration might be caused by the fresh protruding nails that cause more grip in the steel shod condition. It was not possible to calculate the duration of the stance phase, because it was sometimes hardly possible to determine the end of the stance phase on the basis of the accelerometer signal. It is known that raising of the heels preceding toe lift-off sometimes causes an accelerometer signal of a magnitude comparable to that which is evoked at lift-off of the toe[29]. Shoeing the horse with steel shoes increased both the maximal amplitude and the frequency of the vibrations caused by impact, while PU shoes showed a similar power spectrum to that of the unshod condition. This difference between steel shod and unshod was also found in the three horses as reported by Dyhre-Poulsen *et al.*[12]

In conclusion, we can say that PU shoes provide lower-impact accelerations and less friction compared with unshod hooves. Thus, these shoes modify hoof impact in a way that seems promising, given the fact that high-frequency oscillations increase the risk of damage to subchondral bone and joint tissues[8–10].

Table 2 Maximum acceleration data and temporal kinematic variables recorded at trot of the total group of 12 horses (mean ± SD) under the three shoeing conditions

Variable	Unshod	Steel shoe	PU shoe
a_{ymax} (m s^{-2})	504 ± 219	688 ± 293	343* ± 251
a_{xmax} (m s^{-2})	202 ± 124	206 ± 77	105* ± 63
$a_{totalmax}$ (m s^{-2})	540 ± 229	719 ± 287	362* ± 247
t_{impact} (ms)	8* ± 4	14 ± 7	15 ± 7
t_{stride} (ms)	685 ± 32	701* ± 33	688 ± 32

*Statistically significant different values when compared to both other conditions ($P < 0.05$).

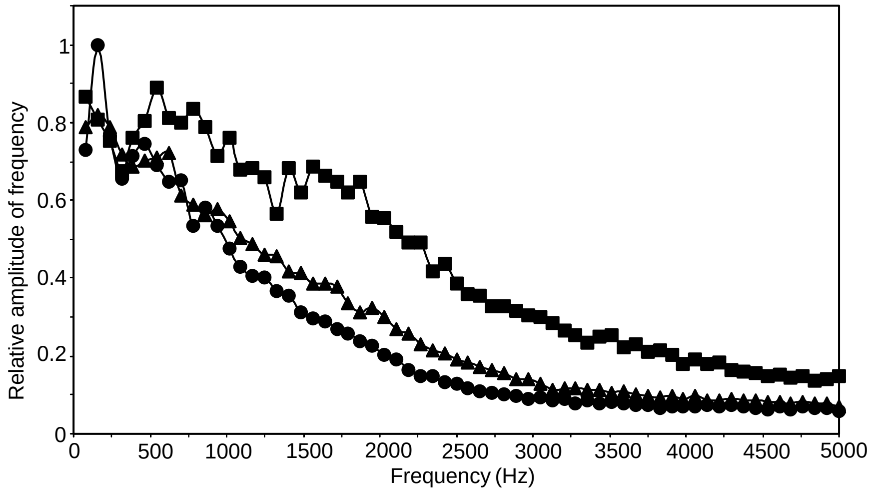


FIG. 5 Fast Fourier transformation curves of impact under the three shoeing conditions: ▲, unshod; ■, steel shoe and ●, PU shoe

Whether steel shoes are deleterious when compared with the unshod condition to the individual sound horse is still a difficult question to answer yet. In general, impact decelerations tend to increase with impact velocity[14] and shorter time of hoof braking leads to higher decelerations and more rapid oscillations[13].

Soft tissues in the hoof act as a low-pass filter, thereby attenuating higher deceleration frequencies; 67% of damping takes place at the hoof wall–distal phalanx interface, while 13% of the initial amplitude remains detectable at the level of the metacarpus[15]. Furthermore, it appeared that shod horses show narrower peaks[12] and a 15% higher amplitude at hoof wall than the unshod horse[15]. Nevertheless, amplitude difference between shod and unshod is absent at the level of the proximal phalanx and distal metacarpus. So, the effect of shoeing at the metacarpophalangeal joint level seems to be minimal[15].

Recent human studies also emphasize that individual differences on impact, in response to different shoes, are caused by varying vertical velocity of the heel at touchdown, plantar ankle flexion and initial knee flexion. Increases in muscular activity might be used by some individuals to move the frequency and damping characteristics of the soft tissues away from those of the impact force and thus minimize vibrations during walking and running, affecting fatigue, comfort, work and performance. Muscle activity is tuned to impact force characteristics to control the soft tissue vibrations in the limb related to impact. Neuromuscular control mechanisms would respond to experiences at one touchdown, change the intensity of muscle activation already before the next heel strike and thus modify next impact, keeping those forces constant, i.e. with different shoe/sole hardness or different surfaces[23].

In horses, Dyhre-Poulsen *et al.*[12] found a 30 ms delay between impact and pressure decline in unshod and 50 ms in shod horses. In general, 30 ms is considered as the minimal neuromuscular response time[2]. Therefore, Lanovaz *et al.*[14] suggested the importance of interphalangeal joints playing a role in passive amplitude attenuation, thus explaining the difference between *in vitro*, *in vivo* and different shoeing conditions. In our study, indeed, horses demonstrated that they could influence their maximal amplitude on impact through modifying their landing technique pre-impact. It is already known from navicular lame horses that force and stress in those horses approximately double the values early in the stance phase of the sound horses group. This is due to a higher force in the deep digital flexor tendon, which is attributed to a contraction of the muscle pre-impact and in early stance in an attempt to unload the heels and thus to diminish palmar heel pain[30], which is in fact a similar mechanism as reported in humans[23].

The outcome of this study, therefore, certainly is encouraging and warrants further investigation into the effects of these kinds of shoe manipulations on critical tissue within the distal limb. In this respect, long-term effects of these shoes will have to be investigated too.

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