The effect of foot imbalance on point of force application in the horse

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Summary

Foot imbalance is believed to be a common cause of musculoskeletal injury in the horse; its biomechanical effects are, however, poorly understood. Wedges (angle 3.7° and 5°) were attached to modified shoes to elevate one aspect of both front feet of Thoroughbred-type horses. The point of force application during weightbearing was determined at trot using a forceplate system. A total of 8 horses were studied with a minimum of 4 providing data for each wedge condition. The results demonstrated that application of a standard steel horse shoe to a balanced foot has minimal effect on the point of force trace through stance. Alteration of mediolateral hoof balance resulted in a displacement of point of force application by about 10 mm in the direction of the wedge throughout stance. Elevation of the heels delayed unloading of the heels and elevation of the toe advanced unloading. Reassessment 24 h after shoeing showed minimal change in the point of force trace. This work demonstrates that a horse is unable to compensate for an acute foot imbalance by redistributing the load under the foot. The higher loads in the elevated region are likely to have a detrimental effect on the hoof structure and horn growth in that part of the hoof.

Introduction

The importance of maintaining a 'balanced' foot in preventing musculoskeletal injury is well recognised (Stashak 1987; Smith and Webbon 1994). Assessment of foot balance is, however, largely subjective in nature and individual farriers apply different assessment criteria and can differ by up to 15 mm in what they define as balanced (M. Cauldwell, personal communication). Foot imbalance is believed to cause selective overload of elements of the skeleton but the elevation required to alter significantly the load distribution under the foot is largely unknown (Stashak 1987; Firth et al. 1988). Foot balance may be altered by either trimming or the application of wedges.

When considering foot ground interaction it is useful to imagine that all the force transferred by the foot is applied at a single theoretical point on the ground surface. This is defined as the point of force (PoF) or point of zero moment (Seeherman et al. 1987; Nigg and Herzog 1994). Forceplates are widely used in equine gait analysis (Silvret et al. 1983; Merkens et al. 1986) and it is possible to calculate the position of the PoF relative to the plate from 8 channel ground reaction force data.

Studies on factors affecting PoF data in the horse are, however, sparse; Seeherman et al. (1987) reported that limb ground reaction force and point of force traces may be used to assess lameness and foot balance in a clinical situation; and that the application of wedges would alter the PoF pattern through the stance. They did not, however, express the positional data relative to the foot. An in vitro limb loading experiment (Colahan et al. 1991) demonstrated that the PoF was located on the medial side of hoof midline and that a 20 mm elevation of the lateral side of the foot resulted in lateral movement of the PoF. They saw no consistent effect with medial, heel or toe wedges. Studies using a forceshoe have also indicated that elevation of one aspect of an equine foot results in an increased load in that region of the foot (Ratzlaff 1988; Ratzlaff et al. 1991).

Point of force data are potentially useful for the dynamic assessment of foot balance. As a prerequisite, normal PoF time traces must be determined and the effects of shoeing and imposed standardised imbalances evaluated. In this study we set out to test the hypothesis that horses show a characteristic alteration in PoF trace with imposed foot imbalance.

Materials and methods

Horses

Eight Thoroughbred-type horses age 2–15 years were used. All horses were judged to have no major locomotor or limb conformation abnormalities on veterinary examination and remained sound with constant bodyweight throughout the study. On experimental days, horses were first exercised for 40 min on a mechanical horsewalker (their usual routine).

The horses' front feet were trimmed to a balanced state (unbroken hoof pastern axis and the solar surface perpendicular to the long axis of the metacarpal bone when held off the ground with a metal T square) and shod with 8 x 19 mm steel shoes. Each shoe was drilled and tapped for wedge attachment. Wedges for imposition of mediolateral imbalance were milled from a second set of shaped steel shoes. Plastic wedges (Stromsholm heel wedges)1 were used to create dorsopalmar imbalance. In both cases, the attachment holes in the wedges were countersunk and the wedges attached using countersunk screws which did not impinge on the sole or protrude from the wedge. The steel wedges induced an alteration in hoof-ground angle of 3.7°, the plastic wedges 5.0°. The hind feet were trimmed but left unshod for the
duration of the experiment. All 8 horses were used to provide data for the comparison of shod vs. unshod; 4 for mediolateral imbalance and the other 4 for dorsopalmar imbalance.

Measurement system

A forceplate track similar to that described by Dow et al. (1991) was used with the plate forming part of a runway of 6 mm thick rubber matting. The 8 channels of output (2 Fx, 2 Fy, 4 Fz) from the forceplate (Kistler forceplate model 9287) were amplified using 8 channel charge amplifiers (Kistler model 9861) and passed through a first order low pass filter (60 dB/octave, corner frequency 360 Hz). The data were then logged at 500 samples/s via a 12 bit AD converter (RT1815A) onto a personal computer using in house software. Point of force conversion formulae, as supplied by Kistler, were applied and PoF data written to disk for subsequent analysis.

All horses were familiar with the forceplate regime and the handlers used. Horses were trotted over the forceplate in a straight line at a constant speed comfortable for the individual horse. A minimum of 6 and up to 10 foot strikes were recorded for each forelimb on each occasion. Data were rejected at time of logging if the horse was judged not to be moving freely at constant speed or if any part of the foot was placed within 50 mm of the edge of the forceplate. Foot position on the plate was determined for analysis of mediolateral foot balance as follows: the walkway 4 to 2 m in front of the plate was sprayed with water and a light dusting of limestone flour applied to the plate surface. A plywood template of the shoe with 3 protruding spikes (drawing pins) was placed on the footprint, a piece of laminated graph paper exactly the same size as the forceplate overlaid and the co-ordinates of the spikes read off. These coordinates were measured in mm relative to axes where y is the direction of motion of the horse, and x measures distance in a perpendicular direction, increasing x being in a medial direction. The plots in this paper hence represent viewing the plate from above for a left foot strike. The x direction was therefore reflected for right foot data. Any observational errors in the coordinate measurements are accounted for in the fitted statistical model described below. The co-ordinates were then used to express the PoF data relative to the foot position. In the case of dorsopalmar wedging the breakpoint over was defined to be the most cranial extent of the PoF. This was assumed to be constant and the PoF data expressed relative to that point. The co-ordinates of the 3 pins were also used to determine and correct for variation in foot orientation on the plate in the case of the mediolateral wedging data.

The horses were forceplate after foot trimming but prior to shoeing and again 2 days after the shoes were applied. Wedges were applied in pairs to generate bilateral foot imbalances and prevent any left-right compensation. Forceplate assessment was undertaken shortly after the wedge was applied (20–40 min) and for the dorsopalmar wedges also 24 h after wedge application. At the end of the experiment a final set of shod, unwedged data were collected.

Data processing and analysis

A general linear model was used to analyse the effect of the different shoeing conditions. The technique is summarised below and discussed in more detail in Ramsay and Silverman (1997). After logging the data were preprocessed using a second in-house programme. This took each point of force record, extracted the data which lay between vertical force reaching 30% of maximum (approximately 1% of total stance duration) and dropping to 8% of peak vertical force (approximately 96% of stance duration). This extraction was undertaken as the first few samples vary widely due to resonance of the forceplate-limb system after foot strike and because PoF data are less accurate at very low vertical forces (Nigg and Herzog 1994). The original 130–150 data points in each run were then reduced by linear interpolation to 100 sets of values for Ax and Ay that were evenly spaced in time and corrected according to position and orientation relative to the y axis as described above. A typical functional observation was therefore a 2 dimensional function of time (X(t), Y(t)) where t varies from 0 to 1 during the stance phase, and X(t) and Y(t) are the co-ordinates of the point of force at time t.

The data set consists of 592 separate runs. There were 8 horses with data for left and right feet. All horses provided data for the shod-unshod comparison with 4 horses receiving mediolateral wedges and 4 toeheel wedges. Data were recorded for toe and heel wedges at 2 timepoints. Finally the wedges were removed and each horse observed again in the shod state. This gives a total of 9 possible conditions, counting the condition of being shod with a normal shoe before the wedges are applied separately from that after the wedges have been removed.

A linear model of the form below was fitted to the data:

\[ X_{ijkl} = \mu + \alpha_{ij} + \beta_k + \epsilon_{ijkl} \]

where all the terms are 2 dimensional functions of t, 0 ≤ t ≤ 1. The suffix ijkI refers to the data collected for the Ith observed curve for side j of horse i under condition k. The function \( \mu(t) \) is the overall mean position. The function \( \alpha_{ij}(t) \) gives the way in which side j of horse i differs from the overall mean and \( \beta_k(t) \) gives the effect of shoeing condition k. The standard assumption is made that these effects combine by addition. Observational, and the variance between successive runs of the same horse with the same shoeing condition are accounted for by the error term \( \epsilon_{ijkl} \).

For any particular curve, suffices (x) and (y) were used to denote the x and y co-ordinates of the vector function. The following identifiability constraints were placed on the various effects, each valid for all i:

\[ \Sigma \alpha_{ij}(t) = 0 \]
\[ \sum_{j=1}^{9} \beta_k(t) = 0 \]

The various effects were then estimated by carrying out a separate general linear model fit for each t and for each of the x and y co-ordinates. Since in practice the data are observed at 100 discrete time points, this means that each \( X_{ijkl} \) corresponds to 2 vectors each of length 100, 1 for the x co-ordinates and 1 for the y co-ordinates. The design matrix relating the expected value of \( X_{ijkl} \) to the various effects is the same for all 200 observed values; and so although the procedure involves the fitting of 200 separate models, considerable economy of effort is possible.

The model can be written as \( X = A \theta + \epsilon \) where X and \( \epsilon \) are both vectors of length 592 each of whose elements is a 2 dimensional function on \([0,1] \). The vector \( \theta \) is a vector of the 26 two-dimensional functions \( \mu(t), \alpha_{ij}(t) \) and \( \beta_k(t) \), and \( A = 592 \times 26 \) design matrix relating the observations X to the effects \( \theta \). The identifiability constraints are incorporated by augmenting
the matrix A by additional rows corresponding to the constraints and by augmenting the data vector X by zeros.

Standard theory of the general linear model then gives as the estimator

$$\hat{\theta} = (A^T A)^{-1} A^T X$$

A plot of the estimated overall 'mean' curve $$\mu(t) = (\mu^x(t), \mu^y(t))$$ may therefore be made. Although the individual observations are somewhat irregular, the overall mean curve is smooth, without any smoothing being incorporated into the procedure.

The general linear model fitted for each co-ordinate at each time point allows the calculation of a residual sum of squares, and hence estimated residual variance curves $$s^x_2(t)$$ and $$s^y_2(t)$$ for the $$x$$ and $$y$$ co-ordinates at each timepoint.

To perform tests of significance an F-ratio statistic was calculated for each value of $$t$$. This allows significance tests for various effects to be carried out for each $$t$$ individually by using standard statistical methods. To gain an overall significance test, the maximum over $$t$$ of the statistic is considered, and its null distribution obtained by simulation. The analysis programmes were written in the statistical package Splus (Splus version 3.2) by the authors and run on a UNIX workstation (SPARC 10).

**Results**

Figure 1 shows a comparison between shod and unshod states; each symbol represents 5% of stance time. Foot strike is followed by loading in the centre of the foot until the last 20% of stance when the PoF moves forward towards the toe until the foot leaves the ground. Shoeing moved the PoF towards the centre of the foot in the early part of stance. The magnitude of the excursion towards the lateral heel was also reduced in the shod state. Standard error was 0.5-1.5 mm, demonstrating that the effect was significant in both $$x$$ and $$y$$ directions in the early part of stance. Visual examination of the shoe wear demonstrated that, as is usual, last foot contact occurred on the lateral aspect of the toe. The unshod hoof sizes were 115-129 mm long and 120-141 mm wide, therefore for most of stance the PoF was just dorsal of the midpoint of the foot. The PoF traces for the shod state at the beginning and end of the study were indistinguishable. These data were therefore pooled to provide a single shod state trace.

The effect of fitting medial and lateral wedges on PoF is demonstrated in Figures 2 and 3. The lateral wedge moved the PoF laterally by about 10 mm during the first 10% of stance and by around 5 mm thereafter. At the end of stance, toe off occurred further towards the lateral side of the foot and hence slightly more caudal. The standard errors (Fig 3) were 0.5-1 mm. The differences were significant beyond the 5% level in the early part of the stride, using a pointwise $$t$$ test and taking note that the degrees of freedom for residual variance are so large that the standard errors may be regarded as known rather than estimated. Lateral wedging had little or no effect on PoF in the $$y$$ plane.

Application of a medial wedge resulted in a medial displacement of the PoF this was greatest at the beginning and end of the stance phase. Again toe off was more caudal than in the unwedged state indicating that toe off occurred on the medial side of the foot. The s.e. values for the medial wedge were greater because fewer samples were recorded in this state.

The effect of application of heel and toe wedges relative to the unwedged state are shown in Figures 4 and 5. Both wedges caused an apparent medial displacement of the PoF in early stance which continued through mid stance for the toe wedge (Fig 5). At breakover there appeared to be a slightly more medial track of the PoF with a toe wedge and a slightly more lateral track with a heel wedge, however this is partly an artifact of the temporal effects described next. The PoF position at 80 to 90% of the analysed stance was displaced cranially with a toe wedge and caudally with a heel wedge. This effect is shown most clearly in Figure 5 where the $$y$$ effect is plotted against time. The
Fig 3: Plot of x and y components of PoF trace against time for shod, medial and lateral conditions. Line styles are as for Figure 2. Pointwise s.e. are given.

Discussion

There are systematic errors of up to 20 mm in the calculated PoF on Kistler forceplates, these are however symmetrical with respect to the plate centre and greatest at the edge of the plate (Bobbert and Schamhardt 1990). In this study, data were discarded if any part of the foot was within 50 mm of the plate edge which would reduce such errors to less than 10 mm in the worst case. Additional, but smaller, errors of up to 3 mm will have occurred in the determination of foot position. The large number of trials considered here, the random location of foot placement on the plate and the statistical model used mean that the end effect would be a small increase in the inter run variance for the medial and lateral wedge data (as only these used absolute measurements of foot position relative to the plate surface). The remaining data sets describe relatively small excursions of the PoF, hence the resultant errors were minimal.

Speed of trot was determined by the horse since it was believed that a trot that was comfortable for each particular horse was paramount in achieving data reproducibility (Dow et al. 1991). The PoF values are not directly attributable to absolute limb vertical force but its distribution between the plate sensors hence no correction for speed was considered appropriate.

The level and mode of forceplate resonance that occurs on impact is variable since it is dependent on where on the plate surface the foot lands. In addition the errors in PoF determination are much greater at low vertical forces (<1000N, Bobbert and Schamhardt 1990). The data recorded at the beginning and end of stance were therefore not accurate representations of the true loading pattern and were disregarded. This made determination of point of first foot contact impossible although an indication can be obtained from the PoF path.

The results presented for mediolateral foot imbalance demonstrate that a wedge will move the point of force towards the elevated side of the foot. This will therefore result in a higher stress in the hoof wall on that side. The clinical practice of wedging a foot where the horn tubules have collapsed is therefore likely to accentuate rather than alleviate that effect.
Whether the elevation of a foot in the acute state is a reasonable representation of a long term foot imbalance is, however, unknown.

It is interesting that the effect of the medial wedge was much more pronounced than that of the lateral wedge. This could indicate that the medial wedge creates discomfort which the horse compensates for by altering its locomotor pattern. Another possible explanation is that when a lateral wedge is applied the horse can compensate by adopting a base wide stance which would return the PoF towards its normal position. Presumably this response is not possible with a medial wedge as it would result in the horse's contralateral limbs interfering during locomotion.

Dorsopalmar wedging resulted in an alteration in the timing of the forward movement of the PoF but had no apparent effect on its speed of movement. It is logical that elevation of the heels would support them at a point in stance when normally they would no longer be in contact with the ground. Hence PoF forward movement would be delayed and vice versa for a toe wedge.

There was no apparent effect of dorsopalmar wedging on the position of the PoF at mid stance demonstrating that heel wedges do not 'unload' the heels.

These results also indicate the importance of medio-lateral foot balance in achieving an even load distribution across the foot. High initial forces are experienced when the foot hits the ground (Dalin and Jeffcott 1994); the so called impact spike. The data presented here demonstrate that, particularly with the medial wedge, initial loading will be concentrated on the elevated side of the limb. During conditions of high limb impact forces, for instance during exercise on hard ground, this localisation of impact force could contribute to the development of musculoskeletal injury (Radin et al. 1982; Barrey et al. 1991). Furthermore, throughout stance the PoF was displaced to one side of normal with both medial and lateral wedges. This generalised paraxial loading would also be likely to have detrimental effects on limb and joint function (Smith and Webbon 1994). The frequency response of the forceplate prevents accurate estimation of the transient forces that occur at foot ground impact and hence whether musculoskeletal damage is induced by loading at impact or during the rest of stance.

In conclusion, application of a standard steel horse shoe has minimal effect on the point of force application through the stance; alteration of mediolateral hoof balance results in a movement of the point of force application in the direction of the wedge; and alteration of dorsopalmar foot balance delays unloading of the heels with elevation of the heels and the reverse for elevation of the toes.

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Manufacturers' addresses

1Stromsholm, Bletchley, Milton Keynes, UK.
2Kistler Instruments Ltd, Alton, Hampshire, UK.
3Analogue Devices Inc., Santa Clara, California, USA.
4Mathsoft Inc., Seattle, Washington, USA.
5Sun Microsystems Inc., California, USA.

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