Motion Capture and 3D Analysis of Equine Locomotion

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Dorothy raised and bred horses in Florida and New Jersey. She maintained a farm for a number of years in Spring Lake, New Jersey. She sold the farm and her horses in the early 1970’s. The sale in no way reduced her affection for the equine population.
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SESSION 1:

3D Digital Kinematics
3-D Kinematics of the Equine Distal Forelimb: Implementation of the JCS on Isolated Limbs and Effects of Asymmetric Loadings

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Digital injuries are one the most frequent cause of lameness in sport and race horses. Analysis of joint movements and circumstances of the locomotion that produce them is helpful for a better understanding of the aetio-pathogenesis of those injuries. It is often hypothesized that the equine digital joints can only undergo movements in the sagittal plane (flexion and extension). However this assumption leaves a gap in the comprehensive understanding of joint behavior because other rotational movements are believed to be of major functional importance. These movements are abduction-adduction (or collateromotion) and axial rotation. They are thought to be involved in the pathogenesis of joint injuries and in the generation of pain. The objective of our study was to develop / adapt a method which allows the quantification of 3-D rotations in the 3 digital joints of the horse. Implementation of this method was first performed on isolated forelimbs (Degueurce et al 2000) and was used to assess the effects of asymmetric loadings on these joint movements.

A joint can be considered as the interface at which relative motion is allowed between two rigid bodies. This motion can be defined using orthogonal frames, rigid with the bones and numerically described with an orientation matrix and quantified using several mathematical methods. We chose to use a Joint Coordinate System (Grood and Suntay 1983), which allows computation of rotations about axis which can be anatomically meaningful at the sacrifice of establishing a reference frame with orthogonal axes (Wu and Cavanagh 1995).

Four isolated forelimbs were used. Trihedrons, made of three axes fitted with kinematic markers, were screwed into each phalanx and aligned with the nominal anatomical axes of the bones. They allowed computation of a local frame associated with each bone. The limbs were subjected to compression with gradually increasing force (from 500 to 6000 N) under a loading machine. Loadings were performed with the foot in a neutral position and with the lateral or medial quarter raised by a 12° wedge. Kinematics of the 3 digital joints was resolved using a "Joint Coordinate System" in which flexion-extension was measured around the medio-lateral axis of the proximal segment, axial rotation around the disto-proximal axis of the distal segment and collateromotion (passive abduction/adduction) around the axis perpendicular to the 2 others.

Raising the lateral part of the hoof induced: lateromotion (5.6°±0.8 at 500 N) and medial rotation (6.5°±0.5 at 500 N) of the distal interphalangeal joint (DIPJ); medial rotation (4.7°±0.5 at 6000 N) and slight lateromotion (less
than 1°) of the proximal interphalangeal joint (PIPJ) and medial rotation
(0.9°±0.2) and lateromotion (2.1°±0.4) of the the metacarpophalangeal joint
(MPJ). The opposite phenomenon was observed with a medial wedge.
Under asymmetric loadings, the digital joints underwent collateromotion in
the direction of the raised part of the foot, whereas axial rotation occurred in
the direction opposite to the raised part of the foot (Chateau et al 2001,
Chateau et al 2002). These results confirm the functional importance of
digital joint movements outside the sagittal plane and demonstrate the
substantial involvement of the proximal interphalangeal joint in the digital
balance. These data are helpful for the identification of biomechanical
factors that may predispose to digital joint injury.

Computation of joint angles was achieved using the principle of the JCS.
The main advantage of this method was to measure joint angles around
anatomically meaningful axes: one axis being linked to the proximal
segment, one other to the distal segment and the third one perpendicular to
the 2 others. However, the choice of the sequence of rotations determined
the axes which were used for the interpretation of the results. We chose the
sequence [F],[C],[R] so that flexion was measured around the medio-lateral
axis of proximal segment and axial rotation around the disto-proximal axis of
the distal segment. This choice was made on the basis of, first, an
experimental study in which the effects of the 6 possible sequences were
studied on our data and, second, on the basis of the recommendation of the
ISB for human biomechanics (Wu and Cavanagh 1995). Interpretation of the
results greatly depends on this choice and it is therefore very important to
precisely describe which convention and axes are chosen when data have
to be compared between different studies. The model was deliberately
limited to 3 degrees of freedom in rotation because rotations are
independent of the location of the local frame; they only depend on
orientation of the frame. On the contrary, magnitude of the translation vector
depends on where the reference points are located in each bone. Unless
these reference points are placed over anatomically meaningful landmarks
(for example, insertion of ligaments), interpretation of translations is most of
the time very confusing.

Main errors were due to the orientation of the axes of the trihedrons. Inter-
limb variability was high for the absolute values of angles (# 4.1° ± 0.9) but
low for the relative (normalized with the neutral test) values (# 0.7° ± 0.3).
Small errors in the orientation of the trihedrons induced imprecision on
absolute values of angles and the neutral test provided reference data to
which other conditions were compared.
This in vitro study was useful to implement the method and to demonstrate
its ability to quantify subtle angular modifications. However, data obtained
on isolated limbs have to be considered with some caution and it was
desirable to try to adapt this method in the living horse.
References


The objective of this study was to measure 3-D rotations of the digital joints in horses walking in a straight line or during a sharp turn. In vitro setup that was previously used on cadaver forelimbs was limited to the analysis of the mid-stance phase and did not reproduce the actual loading of the limb during the whole stance phase. In vitro studies cannot simply be extrapolated to in vivo condition because the rate of loading and orientation of the limb cannot completely mimic the situation in living horses. Those limits justified to find solution for the adaptation of the method in vivo (Chateau et al 2004).

Four healthy French Trotter horses were used. Triads of ultrasonic kinematic markers were surgically linked to the 4 distal segments of the left forelimb of each horse. 3-D coordinates of these markers were recorded in horses walking in a straight line or during a sharp turn in the left direction. 3-D angles of each joint were calculated by use of a Joint Coordinate System as well as the 3-D orientation of the hoof and third metacarpal bone. A calibration procedure was developed to convert data from measurements within a technical coordinate system to data in relation to an anatomically relevant coordinate system. Flexion-extension was measured about the medio-lateral axis of the proximal segment, axial rotation about the disto-proximal axis of the distal segment and collateromotion (passive abduction-adduction) about the floating axis, mutually perpendicular to the other two.

Precision of the method was evaluated to 0.5°, and repeatability of the calibrations resulted in variations that ranged between 0.5° and 1.9°. During the stance phase in a turn, the inside forelimb underwent an adduction (approx 7.3°) that induced lateromotion (approx 2°) and medial rotation (approx 10.2°) in the distal interphalangeal joint and medial rotation (approx 4°) in the proximal interphalangeal joint. These movements were maximal at heel-off and decreased during breakover as the hoof underwent a sudden lateral rotation (Chateau et al 2005).

Interpretation of the results has to be done with careful regard to the methodological choices, first because the sequence of rotations (and therefore axes of interpretation) conditions the results and must be clearly expressed; and second because the method is very sensitive to the definition of these axes. These axes should reproduce as closely as possible the nominal anatomical axes of the bones. However bones are not perfect parallelepipeds and a normal coordinate system can only approximate anatomical axes. We observed that small inaccuracy in the axes definition may shift the absolute value of angles but preserved the
pattern of joint motions. This justified reporting relative values rather than absolute values, each horse being its own control.

An invasive method was used because we wanted to measure the actual movement of the 4 distal segments without resort to skin markers. No horse revealed lameness during the experiment with the only use of analgesics (phenylbutazone, 2 mg/kg, PO, q 12h). However, due to the complexity of the invasive equipment, measurements were limited to slow gaits (walk and slow trot) and further studies are obviously needed for faster gaits. For ethical reasons, it was decided to limit the number of subjects to 4 sound horses. It is therefore acknowledged that extrapolation to lame horses and generalization of the results should be considered with some caution.

Adaptation of the JCS to the digital joints of the moving horse has completed results already obtained on isolated forelimbs and provides new data for the description of movements that were only considered from extrapolation of quasi-static situations. This study illustrates the functional importance of extrasagittal rotations of the digital joints and offers the opportunity to derive hypotheses on biomechanical factors that could contribute to the pathogenesis of digital injuries.

References


3D Kinematics of the Equine Distal Forelimb at Walk and Trot

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A series of studies was performed with the objective of measuring 3D rotations of the digital joints in the equine forelimb during walking and trotting in a straight line on a level surface. Marker triads, rigidly fixed to the bone segments proximal and distal to each joint, were tracked at 120 Hz with an 8 camera Motion Analysis System as the horses walked and trotted along a rubberized runway. Joint kinematics were calculated in an anatomically-based joint coordinate system between the two bone segments using singular-value decomposition (Soderkvist and Wedin 1993) and a spatial attitude method (Woltring 1994).

Distal Interphalangeal Joint (DIPJ)

The DIPJ plays an important role in allowing the solar plane of the hoof to lie flat on the ground, regardless of limb angulation, which implies that the joint has the ability to rotate in directions other than flexion/extension in the sagittal plane. The objectives of this study were to measure 3D rotations of the equine DIP joint in an anatomically relevant BCS during stance and swing phases for horses walking and trotting in a straight line on a level surface.

The BCS for P3 was defined using two markers aligned along the dorsal midline of the hoof wall at the toe and two markers placed equidistant from the toe markers on the medial and lateral quarters. The flexion/extension axis (y) was defined first as the vector running from the lateral quarter to the medial quarter. The internal/external rotation axis (z) was defined as a vector parallel to the vector running from the lower toe marker to the upper toe marker and directed proximally. The adduction/abduction axis (x) was defined as a vector pointing cranially and perpendicular to the plane formed by the flexion/extension axis (y) and the internal/external rotation axis (z). The origin of the BCS for P3 was embedded in the bone midway between the markers at the lateral and medial quarters. It must be pointed out that the internal/external rotation axis (z) and the flexion/extension axis (y) may not be orthogonal if the hoof wall is asymmetrical on the medial and lateral sides.

The BCS for P2 was established using skin markers attached over the proximal tubercles on the medial and lateral sides and the distal tubercle on the lateral side of P2. The tubercles were identified by palpation aided by fluoroscopy. In setting up the BCS, the flexion/extension axis (y) was defined first as the vector running from the proximolateral marker to the proximomedial marker. The adduction/abduction axis (x) was defined as a vector pointing cranially, perpendicular to the plane formed by the flexion/extension axis (y) and the vector running from the proximolateral marker to the distolateral marker. Finally, the internal/external rotation axis (z) was defined as a vector pointing proximally along the long axis of the
bone and perpendicular to the plane formed by the flexion/extension and adduction/abduction axes. The origin of the BCS for P2 was embedded in the bone midway between the two proximal markers.

The patterns of joint motion and angular ranges of motion at the DIPJ were similar at walk and trot with the following ranges of motion:
- flexion/extension: $46 \pm 3^\circ$ at walk, $47 \pm 4^\circ$ at trot; internal/external rotation: $5 \pm 1^\circ$ at walk, $6 \pm 3^\circ$ at trot; and adduction/abduction: $5 \pm 2^\circ$ at walk, $5 \pm 3^\circ$ at trot.

Immediately after ground contact, the DIPJ showed rapid flexion with a little adduction and internal rotation in all horses. The joint then flexed more slowly to reach maximal flexion early in stance. In the second half of stance, extension of the DIPJ was accompanied by a few degrees of internal rotation. Peak extension and internal rotation occurred slightly before lift off in both gaits. In the second half of stance, two horses maintained a little abduction of the DIPJ relative to its angle at ground contact in both walk and trot, whereas the other two horses showed adduction in both gaits at this time.

During the swing phase, rotations of the DIPJ were small, generally changing by less than $3^\circ$ from the values at ground contact. In the sagittal plane, the flexion that began in late stance continued through early swing, returning the joint to its ground contact angle in early swing during walking and by mid-swing during trotting. In the terminal part of the swing phase, all horses showed a small cycle of extension as the hoof was prepared for ground contact.

The flexion/extension axis of the DIPJ as defined by the BCS in this study may not be aligned with the sagittal plane of the horse’s body, depending on the horse’s conformation and the way the limb is placed on the ground. In spite of differences in the analytic planes between 2D and 3D kinematic analyses, the shape of the flexion/extension curves presented here is quite similar to published 2D kinematics of the DIPJ joint at walk (Hodson et al. 2000) and at trot (Clayton et al. 2000). This was not surprising since it has been shown that even large misalignments of the flexion/extension axis have little effect on the results when flexion/extension dominates (Ramakrishnan and Kadaba 1991).

Since the distal limb joints have small dynamic ranges in the dorsal (adduction/abduction) and transverse (internal/external rotation) planes compared with the sagittal (extension/flexion) plane, motions measured in these planes are particularly sensitive to even slightly erroneous marker placement. Sagittal plane angles are also influenced by errors in marker placement, but the shape of the curve may still appear “normal” simply because the signal-to-noise ratio is better for the larger range measurements. This study used a combination of palpation and radiography to place the markers, but these techniques cannot exclude the possibility of some inconsistencies between horses that might contribute to
inter-horse variability. Chateau et al. (2004) used a calibrating device to assist in orienting the local coordinate system relative to the anatomical axes of the bone and reported data for the stance phase of the walk that were similar in all three directions to those presented here.

The ranges of motion outside of the sagittal plane were small in horses moving in a straight line on a level surface. The application of a wedge on the heel, toe, medial or lateral side of the hoof resulted in a smaller angle at the DIPJ on the side to which the wedge was applied (Chateau et al. 2004a, 2006). The ability to move in all directions is likely important in compensating for uneven ground and for angulation of the limb when the horse turns sharply. Chronic hoof imbalance may predispose to osteoarthritis (low ringbone) or soft tissue strain.

Within each gait, kinematic profiles were similar between horses with the exception of adduction/abduction during breakover, which may vary depending on the direction of hoof rotation over the toe. Knowledge of the types and amounts of motion at the DIPJ will be useful in understanding the etiology and formulating treatment strategies for injuries to the soft tissues supporting this joint. These injuries are being recognized more frequently through the use of sensitive imaging techniques.

**Proximal Interphalangeal Joint (PIPJ)**

In kinematic studies, P1 and P2 have usually been modeled as a single rigid segment, which ignores the contribution of the PIPJ to digital motion. The objectives of this study were to measure 3D kinematics of the PIPJ during walking and trotting in a straight line on a level surface.

The BCS for P2 has been described above. The BCS for P1 was established using skin markers attached over the proximal tubercles on the medial and lateral sides and the distal tubercle on the lateral side. The tubercles were identified by palpation aided by fluoroscopy. In setting up the BCS, the flexion/extension axis (y) was defined first as the vector running from the proximolateral marker to the proximomedial marker. The adduction/abduction axis (x) was defined as a vector pointing cranially, perpendicular to the plane formed by the flexion/extension axis (y) and the vector running from the proximolateral marker to the distolateral marker. Finally, the internal/external rotation axis (z) was defined as a vector pointing proximally along the long axis of the bone and perpendicular to the plane formed by the flexion/extension and adduction/abduction axes. The origin of the BCS for P1 was embedded in the bone midway between the two proximal markers.

The ranges of angular motions at the PIPJ show that flexion/extension (range of motion: 13 ± 4° at walk; 14 ± 4° at trot) was the domain rotation. Although the ranges of flexion/extension were similar in stance and swing phases, most of the motion ascribed to the swing phase involved completion of the stance phase cycle of extension. There were small amounts of internal/external rotation (range of motion: 3 ± 1° at walk; 4 ± 1°
at trot) and adduction/abduction (range of motion: 3 ± 1° at walk; 3 ± 1° at trot). All three rotations showed similar ranges of motion in walk and trot. These findings confirm the impressions of Chateau et al. (2004b) that the PIPJ contributes significantly to digital motion in the sagittal and extrasagittal planes.

The flexion/extension patterns were broadly similar for the two gaits when the relatively shorter stance duration of the trot was taken into account. The PIPJ flexed during the impact phase, which is a mechanism for dissipating impact shock as hoof motion is decelerated following ground contact. After the impact phase ends, the PIPJ extends as vertical loading of the limb increases and the body moves forward over the grounded hoof. Extension continues until just before lift off when the joint flexes rapidly through terminal stance and early swing. In the swing phase, three flexion peaks could be distinguished that were consistent in time of occurrence, though their magnitude varied between horses.

In vitro loading of cadaver limbs with a 12° wedge on the medial or lateral side of the hoof caused axial rotation (4.7°) away from the wedge and slight narrowing of the joint (adduction or abduction) on the side of the wedge (Chateau et al. 2002). Compensation for a unilateral heel wedge in vitro was greatest at the DIPJ when loading was low, but at higher loads there was greater involvement of the PIPJ and MCPJ. If this relationship is also true in vivo, it offers an explanation for the high incidence of osteoarthritis of the PIPJ (high ringbone) in horses competing in sports that combine speed with abrupt changes of direction. It also suggests that mediolateral hoof imbalance may predispose to PIPJ osteoarthritis.

The PIPJ also plays a role in compensating for hoof imbalance in the sagittal plane: heel wedges result in increased maximal flexion and decreased maximal extension of the PIPJ, whereas toe wedges are associated with decreased maximal flexion (Chateau et al. 2006).

It is concluded that the PIPJ made a significant contribution to flexion/extension of the digit, but the range of motion did not appear to be strongly influenced by gait or speed. It is questionable whether a joint that shows 14° range of motion in flexion/extension should be regarded as a low motion joint. PIPJ rotations outside of the sagittal plane were small when horses moved in a straight line on a level surface. The findings are relevant in relation to surgical arthrodesis when consideration should be given to the loss of PIPJ extension during late stance, which is likely to be accommodated by increased extension of the DIPJ. For horses working on uneven ground or turning abruptly, the DIPJ and MCPJ may also be required to compensate for loss of axial rotation and adduction/abduction in the arthrodesed PIPJ, which may increase the risk of injury, especially to the collateral ligaments of the DIPJ.

Metacarpophalangeal Joint (MCPJ)
The MCPJ and its supporting soft tissues are common sites of injury in athletic horses. This joint is known to undergo a large range of motion in the sagittal plane, but motions other than flexion/extension have not been studied extensively. The objectives of the study were to measure 3D rotations of the MCPJ during walking and trotting in a straight line on a level surface.

The BCS for P1 has been described above. Skin markers for the MC segment were placed over the dorsal edge of the head of the MC2 and MC4 and distally over the lateral condyle at the attachment of the lateral collateral ligament. For the metacarpal BCS, a right-handed coordinate system was developed by first defining the flexion/extension (y) axis as the vector running from MC4 to MC2. The adduction/abduction axis (x) of the metacarpus was defined as a vector pointing cranially that was perpendicular to the plane formed by the flexion/extension axis (y) and the vector running from MC4 to the lateral metacarpal condyle. Finally, the internal/external rotation axis (z) was defined as a vector pointing proximally along the long axis of the MC, perpendicular to the plane formed by the flexion/extension and adduction/abduction axes.

The ranges of motion of the MCPJ were: flexion/extension: 62 ± 7° at walk, 77 ± 5° at trot; adduction/abduction: 13 ± 7° at walk, 18 ± 7° at trot; and axial rotation: 6 ± 3° at walk, 9 ± 5° at trot. Unlike the DIPJ and PIPJ, the range of flexion/extension at the MCPJ was greater at trot than at walk, which is in accordance with other studies that have reported a correlation between peak extension and speed (McGuigan and Wilson 2003).

MCPJ lesions, such as chip fractures and osteoarthritic changes, frequently affect the dorsal margin of P1, which corresponds with a dorsal shift in the contact area of the MCP joint as speed increases (Brama et al. 2001). These changes in contact area are a consequence of the increase in stance phase extension of the joint as it is more heavily loaded.

Flexion/extension had a consistent pattern and amplitude in all horses and appeared to be coupled with adduction/abduction, such that stance phase extension was accompanied by abduction and swing phase flexion was accompanied by adduction. Though it is possible that there was some kinematic cross talk between flexion/extension and abduction/adduction. Metacarpal condylar fractures occur almost exclusively in horses galloping at racing speed, which is associated with marked extension of the metacarpophalangeal joint during weight-bearing (Parkin et al. 2005). The predominance of fractures of the lateral condyle, which account for 85% of condylar fractures (Zekas et al. 1999), may be a consequence of greater compressive forces on the lateral side of the MCPJ as a consequence of compression of the lateral side of the MCPJ as it is loaded in extension and abduction.

It is concluded that flexion/extension is coupled with adduction/abduction at the MCPJ and that loading of the joint in extension and abduction offer an
explanation for the development of injuries to the dorsal margin of P1 and the lateral MC condyle.

References

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Reliability of a Clinically Practical Multi-segment Foot Marker Set / Model

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Improvements in image resolution within the digital camera industry have dramatically increased our ability to accurately track small markers and small marker clusters. These abilities have generated substantial interest in improving aspects of gait analysis associated with measuring the kinematics and kinetics of foot motion. As a result of this interest, numerous multi-segment foot marker sets have been proposed for use in clinic and research. These models differ in the number of segments used to define the foot, the anatomical structures used to define the segments, and the manner in which segments are represented mathematically. However, a general problem needs to be addressed prior to the implementation of any of the proposed multi-marker foot models: the ability of clinicians to accurately implement the model. As markers are placed in closer proximity to other markers, the precision of marker placement becomes more critical to the validity of the results. As markers are placed closer together, small errors in placement have a larger effect on the orientation of segments defined by the markers. Marker reliability measures were obtained using 14 adult feet (mean=30 ± 7.8 yrs, average foot length=246.26mm ± 22.11mm) and 8 pediatric feet (mean=6.5 ± 2.6 yrs, mean foot length=196.38mm ± 28.80mm). Three clinicians with extensive experience in gait analysis identified anatomical landmarks on the foot necessary to produce the desired 4-segment foot model. The first part of the analysis examined the absolute marker positions within marker application, between marker applications, and between clinicians. Our analyses indicate that the precision of marker placement at these specific anatomical locations on the foot is limited. The motion capture system produced less than 1mm error for markers placed at least 1cm apart, which means that this was not a substantial source of error. We found good reliability of repeatedly placing markers on the foot by a single clinician, but poor reliability when multiple clinicians were involved. As current camera systems are able to detect smaller markers placed in close proximity, the problem shifts from limitations of technology to clinical practicalities, and the limiting factor becomes the ability to accurately and reliably place markers on the foot.
SESSION 2:

Hoof Motion and Effects of Shoeing

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It is common practice for equine veterinarians and farriers to use shoeing and hoof trimming techniques to attempt to change hoof orientation during the different phases of the stride. Changing hoof orientation is thought to affect internal hoof biomechanics, which spares anatomic structures within the foot and distal limb from excessive load or strain. Also, redirecting the direction of hoof breakover is, in essence, changing hoof orientation at the end of stance, and this is desired frequently to alter hoof flight during the swing phase of the stride to eliminate or reduce interference with the opposite limb. However, there is little objective supportive evidence that the techniques employed by veterinarians and farriers for changing hoof orientation are, in fact, effective.

One of the main problems of gathering objective evidence of hoof orientation in the horse during the stride is handling the expected natural stride-to-stride variability. The horse is not a machine and body position, limb swing, hoof placement, etc., should vary quite considerably during nature motion. Therefore, in able to detect small changes in hoof orientation between treatments or conditions, which could be clinically important, many strides must be analyzed. Outside the treadmill laboratory this is difficult to accomplish with video-based kinematic gait analysis or with tethered body sensor kinematic measurements.

Here we describe a completely wireless 3D gyroscopic system for measurement of hoof orientation in the horse. The system was used to test whether or not a commonly used shoeing technique in horses affects the direction of breakover. The system could be equally well used to field test other shoeing or trimming techniques in the horse. We also describe an associated and user-friendly, data analysis, animation, statistics generating and data-base storage program.

Tranducers, which are attached to the dorsal hoof wall, consist of 3 Gyrostar® piezoelectric vibrating ENC series gyroscopic sensors (Murata Manufacturing Company, Ltd, Kyoto, Japan), a Class 1 Bluetooth (or Zigbee) Wireless Module, a 4.5 V, 250 mA lithium polymer battery and a PIC 18F4520 microcomputer in a non-inverting, low-pass-filter, integrated circuit design. A transceiver, which is either attached to the horse’s back with a girth strap or held by a handler with the horse, consists of a Class 1 Bluetooth (or Zigbee) Wireless Module and a PHS (Personal Handyphone System) cell phone. The data is then transmitted to a PHS-enabled PCI card in a laptop computer. Data collection is accomplished in real time.
Data analysis, hoof model animation data archiving are performed subsequent to data collection.

Signals of rotational angular velocity for pitch, roll, and yaw of the hoof are converted to rotational angular position of pitch, roll, and yaw by integration. Rotational angle of the hoof is calibrated to zero by subtracting the most frequent number within the signal (ie, statistical mode), from the raw signal values. Rotation of the hoof in one direction is defined as a positive deflection and rotation in the opposite direction is defined as a negative deflection.

Start and end of breakover for each stride is defined from the rotational pitch channel. Breakover begins at the first negative deflection of hoof pitch and ends when hoof pitch reached a minimum negative position. Beginning and end of breakover are determined for each stride. Total breakover duration is then divided into 4 equal segments. Output that determines breakover direction are the roll and yaw angles at the end of each segment of breakover for each stride.
The Effects of Heel and Toe Elevation on the 3-D Kinematics of the Hoof and Digital Joints

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Comprehensive understanding of the three-dimensional (3D) kinematics of the distal forelimb and precise knowledge of alterations induced by dorsopalmar foot imbalance remains incomplete because in vivo studies performed with skin markers do not measure the actual movements of the three digital joints. The objective of this study was to quantify the effects of 6° heel or toe wedges on the 3D movements of the four distal segments of the forelimb in horses trotting on a treadmill.

Three healthy horses were used. They were equipped with ultrasonic markers surgically fixed to the four distal segments of the left forelimb. The 3D movements of these segments were recorded while horses were trotting on a treadmill. Rotations of the digital joints were calculated by use of a “Joint Coordinate System”. Data obtained with 6° heel or toe wedges were compared to those obtained with flat standard shoes.

Use of heel wedges significantly increased maximal flexion of the proximal (PIPJ) and distal (DIPJ) interphalangeal joints and decreased maximal extension of the PIPJ and DIPJ. Inverse effects (except for PIPJ maximal extension) were observed with the toe wedges. In both cases, neither flexion-extension of the metacarpophalangeal joint nor extrasagittal motions of the digital joints were statistically different between conditions.

At a slow trot on a treadmill, heel and toe wedges affect the sagittal plane kinematics of the interphalangeal joints. Better understanding of the actual effects of toe and heel wedges on the 3D kinematics of the three digital joints may help to improve the clinical use of sagittal alteration of hoof balance in the treatment of distal forelimb injuries.
The Effects of Egg-Bar Shoes on the 3-D Kinematics of the Hoof and Digital Joints

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Understanding of the biomechanical effects of egg-bar shoes remains incomplete because kinematics studies are usually performed on hard tracks and with skin markers which do not measure the actual three-dimensional (3D) movements of the three digital joints. The objective of this study was to quantify the effects of egg-bar shoes on the 3D kinematics of the distal forelimb in horses walking on a sand track.

Four healthy horses were equipped with ultrasonic markers surgically fixed to the four distal segments of the left forelimb. The 3D movements of these segments were recorded while the horses were walking on a sand track. Rotations of the digital joints were calculated by use of a Joint Coordinate System. Data obtained with egg-bar shoes were compared to those obtained with standard shoes. Mean differences were expressed in a 0.95 confidence interval.

With egg-bar shoes, the initial sinking of the heels into the ground during landing was reduced and the heels were raised by up to 5.1° (3.5-6.7°) compared to standard shoes at mid-stance. Concurrently, maximal flexion of the distal (DIPJ) and proximal (PIPJ) interphalangeal joints was increased by up to respectively 3.2° (2.2-4.2°) and 1.8° (1.1-2.5°) at the beginning of the stance phase. At heel-off, extension of the DIPJ was reduced by 3.8° (2.6-5.0°). In extrasagittal planes of movement, egg-bar shoes prevented sinking of the medial quarter into the ground which led to a slight decrease of DIPJ medial rotation and lateromotion.

Egg-bar shoes prevent the heels and, to a lesser extent, the medial side of the hoof from sinking into the ground on a sand track. They contribute to a decrease of DIPJ maximal extension at heel-off and to hoof stabilisation in the transversal plane.

Such quantitative results are useful to support the clinical indications of egg-bar shoes in the horse.
Joint Coordinate Systems for 3-D Movement Analysis

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Introduction
Three dimensional movement analysis can produce clinically relevant information on joint kinematics (movement) and joint kinetics (forces). Here we will review how joint coordinate systems are used to obtain standardized quantitative descriptions of joint kinematics and kinetics.

History
In two dimensional movement, joint rotation can be described by a single joint angle. Joint kinetics is typically presented as a resultant force vector, and resultant joint moment which represents the action of muscles. In three dimensions, joint rotation can be described as an ordered sequence of three rotations, also known as the method of Euler angles. In (aero)nautical engineering, there is a well established interpretation and terminology (roll, pitch, yaw). An early adopter of this method was Crowninshield in his studies on hip joint kinematics and kinetics [2]. However, Crowninshield’s main interest was in estimating articular contact forces, in which the kinematic analysis was only an intermediate step. Around the same time, Grood and Suntay [4] worked on quantification of joint motion and were concerned that the Euler angle method was not useful to clinicians. They also felt that a description of joint motion should not require a specification of the order of rotations. Hence, they proposed the use of a Joint Coordinate System (JCS), in which the first axis is attached to the first bone, the third axis is attached to the second bone, and the second axis is a floating axis which is mutually perpendicular to the other two. On each axis, a rotation and translation is defined, and all rotations and translations take place at the same time, as in a mechanical linkage, thus eliminating the need for specifying the order of rotations. It is now understood that this method is identical to the Euler angle method. Nevertheless, Grood and Suntay’s work led to broad acceptance of these engineering methods by establishing a strong link with clinical terminology. For the knee joint, they defined a standard coordinate system such that the three axes were consistent with clinical definitions of flexion, abduction, and rotation, respectively. Similar standards for other joints were subsequently developed [1][8].

Kinematics
When using optical motion capture, kinematic analysis is typically performed in three steps. First, coordinate systems are defined on each bone. Second, the rotation matrix and translation vector relating the two coordinate systems is obtained from 3-D marker coordinates during movement. Finally, Grood and Suntay’s matrix equations are used to extract the three rotation angles and three translations. Computer code for this procedure is readily available in the public domain (http://www.isbweb.org/software/movanal.html). In our own work, we
perform the last two steps simultaneously using the Mocap Solver technology (Motion Analysis Corp., Santa Rosa, CA). Briefly, the limb is modeled as a mechanical linkage which represent the physical equivalent of a joint coordinate system. Markers are defined on the model, and joint rotations and, if needed, translations are solved iteratively from measured marker trajectories. These methods were developed simultaneously in several research groups, and are collectively known as “global optimization”.

**Kinetics**
After adding information on mass properties and external forces, the resultant joint force and moment vectors can be solved from the equations of motion of a linked segment model. The current practice is that resultant joint force and moment are expressed in the coordinate system of one bone, not in a joint coordinate system. This has some practical advantages, for instance, hip joint kinetics can be obtained without capturing the motion of the pelvis. Also, further analysis to obtain articular contact forces is more relevant in a bone coordinate system, such that the results can be directly used for mechanical testing of orthopaedic implants. For quantification of muscle function, however, one could make the case that joint moments should be quantified on the same axes as the joint motion. When large rotations occur in the joint, the results can be quite different.

**Common problems**
JCS rotation angles become increasingly sensitive to measuring errors when the second rotation approaches ±90 degrees [7]. This is the well known “gimbal lock” problem, which does not occur in the knee joint, because the ab-adduction motions remain well below 90 degrees. However, this issue is a major obstacle in quantification of shoulder rotations. For analysis of baseball pitching, we have adopted the “globe method” in which the first axis of the JCS is the inferior-superior axis of the thorax [3]. This works well during the throwing motion, but the gimbal lock problem now occurs in the neutral pose. Screw axis descriptions of joint motion have been proposed [7], but these remain difficult to translate into clinical terminology.

Another known problem is crosstalk, which is commonly seen at the knee. A small misalignment of the flexion axis will result in a small part of the flexion movement being misinterpreted as abduction. Because the true abduction is small, this causes a large relative error [5]. This problem can be minimized by carefully deciding how to define the JCS axes. Axes based on anatomical landmarks are not always sufficiently reproducible. When this problem occurs, one should also consider whether such a small rotation is a useful variable. Apparently the joint is very stiff in that axis, and joint moment may be more relevant and reliable than joint rotation. In Mocap Solver, such ‘unused’ degrees of freedom can be removed from the model in order to obtain more robust results. For the ankle joint, it is beneficial to use a reduced joint coordinate system that represents two functional axes, rather than three anatomical axes [6].
## References

SESSION 3:

3D Kinematics of Proximal Limb (horse and human)
3D Kinematics of the Equine Tarsal Joint at Trot

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The tarsal joint complex is frequently a site of pathology, including degenerative lesions of the low motion joints in the distal tarsus. The majority of tarsal joint motion occurs at the tarsocrural joint, where the obliquity of the trochlear surface of the talus is well known. As a consequence of this obliquity, the tarsocrural joint is said to have a helical or screw action. The objectives of this study were to characterize the six DOF motion of the equine tarsus by measuring motion of the MT3 relative to the tibia during normal trotting and to test the hypothesis that tarsal joint action is exclusively due to the screw action of the tarsocrural joint.

Steinmann pins were inserted into the lateral shaft of the tibia and MT3 under general anesthesia. Triads of reflective markers were attached rigidly to each bone segment during collection of kinematic data. Prior to collection of data at trot, skin markers were attached over each segment to establish the bone coordinate systems (BCS). For the tibia, the three landmarks were the fibular head at the site of attachment of the lateral collateral femoro-tibial ligament, the lateral malleolus and the medial malleolus. A right-handed coordinate system was developed for the tibia by first defining the flexion/extension axis as the vector running from lateral malleolus to medial malleolus. The adduction/abduction axis of the tibia was defined as a vector pointing cranially and perpendicular to the plane formed by the flexion/extension axis and the vector running from the lateral malleolus to the fibular head. Finally, the internal/external rotation axis of the tibia was defined as a vector pointing proximally and perpendicular to the other two axes, forming a right-handed coordinate system. The origin of the tibial BCS was embedded in the bone midway between lateral and medial malleoli.

For the MT3, the bony landmarks were located at the distal condyle at the metatarsal attachment of the lateral collateral ligament of the metatarsophalangeal joint, the dorsal edge of the head of the MT4, and the dorsal edge of the head of MT2. The BCS for MT3 was defined in a similar manner to the tibial system. The flexion/extension axis was defined as a vector running from MT4 to MT2. The adduction/abduction axis was defined as a vector pointing cranially and perpendicular to the plane formed by the flexion/extension axis and the vector running from MT4 to the distal condyle. The internal/external rotation axis was defined as pointing proximally and perpendicular to the other two axes, forming a right-handed coordinate system. The origin of the third metatarsal BCS was embedded in the bone midway between MT4 and MT2.

Standing files were recorded with both the marker triads and the skin markers in place. From each BCS, a transformation matrix was developed which was used to transform the locations of the triad markers in the
standing position from the GCS to the corresponding BCS. For the trotting trials, the bone-fixed markers were tracked at 120 Hz using a 6-camera Motion Analysis System and marker locations were transposed to the BCS using a singular-value decomposition method (Söderkvist and Wedin 1993) was used calculate the position and orientation of each BCS using the standing and trotting triad marker locations.

The three rotations were expressed using an attitude vector based on the finite helical angle method. The relative angular motion between the tibia and third metatarsus was expressed in terms of a spatial attitude vector (Woltring 1994). This method is based on the finite helical axis representation of relative motion, where the position and orientation of one body with respect to another at any given moment can be described as a scalar rotation around, and a translation along an axis between the bodies.

The angles were expressed as though the tibia (proximal segment) was fixed and MT3 was moving relative to it. Therefore, the flexion(-)/extension(+), adduction(+)/abduction(-), and internal(+)/external(-) rotation components of the attitude vector that describe the angular motion of the tarsal joint are expressed in terms of the corresponding axes in the tibia. Motion along the adduction/abduction axis of the metatarsus was labeled as cranial(+)/caudal(-) translation. Translation along the flexion/extension axis of the metatarsus was labeled as medial(+)/lateral(-) motion. Finally, motion along the internal/external rotation axis of the metatarsus was considered as proximal(+)/distal(-) translation. Translations were expressed as movements of the tibia relative to the MT3 because the BCS of MT3 is more closely aligned with the sagittal plane of the horse.

Data were collected at 120 Hz for four trials per horse. In all horses, the tarsus showed a cycle of flexion then extension during stance, followed by a larger magnitude cycle of flexion then extension during swing. For the most part, the other two rotational degrees of freedom and the translational motions were coupled with the flexion/extension angle, which is what should result from a screw joint motion.

The angular motions (fig 1) are, for the most part, coupled, which would be expected given the screw-like motion of the tarsocrural joint (fig 1). There is a strong coupling between flexion/extension and adduction/abduction angles during the entire stride, with increasing abduction as flexion increases. The flexion/extension and internal/external rotation angles are also strongly coupled during most of swing, showing increasing external rotation with increasing flexion. During mid-swing, there is a period of decoupling which appears in varying amounts for each subject. In the stance phase, the direction of internal/external rotation is opposite to that during swing. There is also a lesser degree of coupling between these rotations during the stance, which indicates that the joint exhibits motion during weight bearing that is not predicted by only the screw motion of tarsocrural joint (Lanovaz et al. 2002).
The three translations are also strongly correlated to the flexion/extension angle (fig 2). The tibial BCS origin traces a 3D circular path which generally resembles the shape of the talar trochlea. The magnitudes of the displacement are consistent with what might be predicted using the anatomical measurements of the trochlea by Badoux (1987), but the translations are not completely proportional to the flexion/extension angles. The most apparent deviation is craniocaudal displacement during swing. At approximately 55% of the stride (25% of the swing phase), the cranial/caudal displacement “flattens” and becomes asymmetric with respect to the swing phase flexion peak. This produces hysteresis when the translations are plotted with respect to flexion/extension angle. One explanation for this behavior is the superposition of some caudal motion of the tibia/talus with respect to MT3 outside of the tarsocrural joint during mid swing. There is evidence of similar caudal translations, although to a smaller degree, in stance, as well as some medial translation during mid stance and mid swing. Since this study only measures motion of the tibia relative to MT3, it is not possible to determine the amount and type of motion occurring at the individual joints within the tarsal joint complex. However, given the assumption of circular talar ridges (Badoux 1984), even conservative estimations indicate significant movement at the other tarsal joints.

It was concluded that, although the majority of tarsal motion occurs in the tarsocrural joint, there is evidence that translations and rotations occur in other locations within the tarsal joint.

References


The complex structure of the equine carpus, with multiple rows of bones and articulations, suggests the likelihood of motions other than flexion/extension in a sagittal plane. Indeed, abduction/adduction and axial rotations can be observed, especially in gaited breeds, such as the Peruvian Paso in which *termino* is a feature of the gait. A study of 3D carpal rotations in Missouri Foxtrotters showed that the flat walk and the fox trot had similar profiles for flexion and extension, but adduction/abduction and axial rotation showed greater differences between gaits (Nicodemus 2000). The objective of this study was to measure 3D carpal joint motion during trotting.

Triads of reflective markers were rigidly attached to the radius and third metacarpal bone (MC3) of the right forelimb at sites where the bone is subcutaneous: on the lateral side of the radius, proximal to the lateral styloid process and on the dorsolateral aspect of the MC3. In addition to the bone-fixed markers, reflective markers were attached to the skin to define the anatomically based BCS for the radius and MC3. Three easily palpable landmarks on each bone were chosen to define the BCS: the radial tuberosity (Radprox), the lateral styloid process (Radlat), the medial styloid process (Radmed), the dorsal edge of the head of the 2nd metacarpal bone (MCmed), the dorsal edge of the head of the 4th metacarpal bone (MClat) and the MC3 condyle at the proximal attachment of the lateral collateral ligament of the metacarpophalangeal joint (MCdist).

Kinematic data of the standing horse were collected with both the bone marker triads and the skin markers in place and, from these files, the radial and metacarpal BCS were established. Data were collected at trot in the laboratory GCS using a 6-camera Motion Analysis System recording at 120 Hz. A successful trial consisted of a single stride starting at ground contact of the right forelimb. Five trials from each horse were recorded, with forward velocities being closely matched between subjects.

For the radial BCS, a right-handed coordinate system was developed by first defining the flexion/extension axis (y) as the vector running from Radlat to Radmed. The adduction/abduction axis (x) of the radius was defined as a vector pointingcranially and perpendicular to the plane formed by the flexion/extension axis (y) and the vector running from Radlat to Radprox. Finally, the internal/external rotation axis (z) of the radius was defined as a vector pointing proximally along the long axis of the bone and perpendicular to the plane formed by the flexion/extension and adduction/abduction axes. The origin of the radial BCS was embedded in the bone midway between Radlat and Radmed.

For the metacarpal BCS, a right-handed coordinate system was developed by first defining the flexion/extension axis (y) as the vector running from MClat
to MCmed. The adduction/abduction axis (x) of the metacarpus was defined as a vector pointing cranially and perpendicular to the plane formed by the flexion/extension axis (y) and the vector running from MClat to MCdist. Finally, the internal/external rotation axis (z) of the metacarpus was defined as a vector pointing proximally along the long axis of the bone and perpendicular to the plane formed by the flexion/extension and adduction/abduction axes. The origin of the metacarpal BCS was embedded in the bone midway between MClat and MCmed.

To calculate the kinematics of joints in the sense of anatomic position, the radial and metacarpal marker triads in the standing file were transformed from the GCS to the corresponding BCS (Grood and Suntay 1983). The orientation matrices and displacement vectors were then calculated for the standing file and for each frame of trotting data using a singular value decomposition method (Söderkvist and Wedin 1993).

Relative angular motion between the radius and MC3 was expressed in terms of a spatial attitude vector (Woltring 1994) calculated with respect to the proximal (radial) segment, so the angular values can be thought of as though the radius being fixed and MC3 moving relative to it.

Within each individual horse, the rotational and linear movements were quite consistent.. In general, the patterns of the rotations and linear displacements were similar but the magnitudes varied between horses, especially for flexion/extension and proximal/distal displacement. The shape and direction of adduction/abduction and mediolateral displacement differed between horses (Fig 1).

Consistent features of carpal kinematics were that, in early stance, the carpus extended and rotated inward, and the origin of the metacarpal BCS moved cranially relative to the radius. This position was maintained until late stance. Just before lift off, the carpus began to flex and rotate externally. Flexion continued into midswing, resulting in proximal and caudal displacement of the origin of the metacarpal BCS. The range of motion for flexion/extension was 15±6° in stance and 76±13° in swing. Differences between horses in range of flexion/extension during stance (Table 1) are primarily due to different amounts of carpal flexion at lift off, rather than to differences in the amount of carpal extension in the earlier part of stance, which was fairly similar in all subjects.

Correlation between flexion/extension versus proximal/distal and cranial/caudal displacements showed some hysteresis around the time of peak flexion (Fig 2), so the curves were analyzed in two parts: from the start of swing to maximal carpal flexion and from maximal flexion to the end of swing. In each case the relationship was primarily linear, but a 2nd order polynomial improved the fit and resulted in an r^2>0.99 in all cases (equations listed in fig 2).

References


<table>
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<th>Kinematic Variable (unit)</th>
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<td>Prox/dist trans (mm)</td>
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<td>5±2</td>
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Table 1: Range of carpal joint rotations and translations in three subjects, five strides per subject, and mean values for the group. Values are mean ± SD.
Figure 1: Carpal joint rotations (left) and translations (right). The lines show mean values for individual horses and mean trace for the group (thick black line) during one stride at trot commencing at hoof contact. The dashed vertical line indicates lift off. Note different scales.
Fig 4: Relationships between flexion/extension and proximal/distal translation (black line) and cranial/caudal translation (grey line). Direction of curves is indicated by arrows.

The equations were as follows:
Flexion/extension versus proximal/distal translation:
Ascending curve: \( y = 0.0055x^2 + 0.1788x - 0.6584 \) \( r^2 = 0.9993 \)
Descending curve: \( y = 0.006x^2 + 0.0711x + 0.8577 \) \( r^2 = 0.9943 \)

Flexion/extension versus translation cranial/caudal
Descending curve: \( y = 0.0052x^2 - 1.3089x - 0.5254 \) \( r^2 = 0.9998 \)
Ascending curve: \( y = 0.0052x^2 - 1.2809x - 0.4739 \) \( r^2 = 0.9991 \)
Clinical Usefulness of Four Functional Knee Axis Algorithms

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There is growing clinical concern regarding the accuracy and repeatability associated with the traditional method of manually identifying the knee flexion/extension axis. Accordingly, investigators have proposed functional methods for defining a single representative knee axis. To date, there has been no report of how these newer methods perform clinically or how they compare to a reliable anatomical reference. Four original functional knee axis algorithms and two algorithm modifications were evaluated with gait data gathered from four normal subjects. Ultrasound measures of tibial torsion were gathered for each subject and compared to dynamic shank rotation profiles determined from each knee axis algorithm. Additionally, medial and lateral knee markers were progressively rotated around the knee of one subject to assess algorithmic sensitivity to original marker placement. Group results showed that the traditional method produced knee axes that were rotated 6° on average from the measures reported using Ultrasound. The knee ab/adduction minimization and helical axis methods produced the most anatomically accurate and consistent functional knee axes with absolute torsion/rotation differences of approximately 3°. Modifications of other algorithms showed little or no improvement over the original methods. Algorithm performance characteristics remained intact when marker placement was intentionally perturbed. Functional knee axis determination using ab/adduction minimization or helical axis estimation provides a more robust alternative to traditional knee marker placement provided patients display normal knee motion. For patients with excessive lateral knee laxity and/or diminished knee flexion range, additional trials of passive knee flexion through a larger range are recommended.
Musculo-skeletal Modeling for Equine Motion Analysis

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Motion capture allows the study of kinematics. The potential of inverse dynamics models to combine these results with kinetics have been widely demonstrated, although the propagation of measurement errors within a model continues to be a problem. Increasingly a middle ground of kinematic models is beginning to show its value in calculating soft tissue strains and forces. Using musculo-skeletal modeling allows the calculation of in-vivo strains and forces without the use of invasive transducers or the requirement to measure ground reaction forces.

Scapula-thoracic Modeling
Ongoing work is looking into the transmission of forces from the thoracic limbs through the scapula to the thorax, and its role in the aetiology of exercise induced pulmonary haemorrhage (EPIH). Muscle modeling can be used to facilitate the non-invasive tracking of the scapula.

Distal Limb Modeling

![Figure 1 Equine Distal Limb Model](image)

A subject-specific equine model has been created which takes inputs from kinematic data and creates a 3D animation of the limb and calculations of tendon and ligament strain.

The shapes of the bones were taken from CT scans of the subject and the tendon insertions and paths from dissection studies. The superficial digital flexor tendon (SDFT) and deep digital flexor tendon (DDFT) took virtual origins at the proximal end of the metacarpus. The suspensory ligament (SL) was modeled from its metacarpal insertion to the proximal sesamoid bones.

The movement of the metacarpus and phalanges were captured for the same horse (Chateau et al., 2004). The movement of the sesamoid bones was simulated using the constraints of maintaining an isometric virtual ligament and maintaining contact between the appropriate articular
surfaces. The simulation of the proximal sesamoid bones was compared to
movement recorded in-vitro and shown to be reliable.

**Tendon Strain**
The influence of muscle bellies on the tendon strain has not yet been
included in the model however the strains calculated are in accord with
those previously published (e.g. Pourcelot et al., 2005; Riemersma et al.,
1996).
Sensitivity
The paths and origins used for the DDFT, SDFT and SL were altered and the effects on their calculated strains during trot stance were examined. Despite the multi-articular nature of the flexor tendons, their paths and moment arms are more sensitive to perturbations in modeling of the sesamoid bones than to the origin points chosen or the action of other joints.

Moments Arms
The moment arm of the equine flexor tendons at the metacarpo-phalangeal joint varies with joint angle, and it has been suggested that a fixed radius pulley such as a sphere is not a sufficiently accurate model for it (Brown et al., 2003). The inclusion of mobile sesamoid bones and suitable wrapping algorithms combined produce a variable moment arm in the ranges of those previously published (Brown et al., 2003; Meershoek et al., 2001). The influence of the distal interphalangeal joints on the tendon path and moment was ignored in previously published results but was clearly visible in the difference between the early and late stance results obtained.

Effect of Toe and Heel Wedges
The sagittal alteration of hoof balance is a common intervention in horses. The effect of toe or heel elevation on tendon strains was investigated using the previously collected data (Chateau et al., 2004). The DDFT showed the same results for walk and trot with the heel wedge decreasing peak strain and the toe wedge increasing it. The opposite results were seen in the SL with the heel wedge decreasing peak strain and the toe wedge increasing it. The SDFT showed similar results to the SL, with the heel wedge increasing peak strain at walk and toe wedge decreasing peak strain at walk and trot.

Influence of Proximal Interphalangeal Joint
As the proximal interphalangeal joint is often ignored in strain calculations its influence on tendon and ligament strains was also tested. The proximal interphalangeal joint was shown to be highly influential increasing the strains calculated with normal shoes and inverting the effect of the wedges.

Collateral Ligaments
Collateral ligament injuries are common at the digital joints in horses. The effect of a sharp turn and a higher speed gait was calculated using the data collected invasively in-vivo. As expected, the effect of the different activities varied between the different branches and attachment sites of the ligaments. In the metacarpophalangeal collateral ligaments the turn introduced the greatest increase in peak strain, although the increased speed also increased strain and loading rate. In the distal interphalangeal joint the sharp turn induced the greatest adduction and medial rotation, and accordingly the peak strain increased in both collateral ligaments. The alignment of the collateral ligaments makes the distal interphalangeal joint more tolerant of movements outside the sagittal plane, whilst restricting the metacarpophalangeal joint to more planar motion.
Increasing Applicability
Future work aims to include the muscle bellies of the DDFT and SDFT in the distal limb model. In order to make the models more widely applicable customisation and scaling routines are being developed as part of developmental and variability studies in young horses and different breeds. The development of joint constraints and skin movement algorithms will also help to reduce the measurement errors of non-invasive markers.

References
A Tutorial on Joint Orientation Measurements with Application to the Human Shoulder

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When evaluating joint motion in clinical environments, the measurement objective is to provide clinicians with: 1) anatomically meaningful descriptions of joint position, and 2) a clinically relevant sense of joint motion. When referring to the shoulder, joint motion usually refers to some combination of scapula/clavicle motion relative to the trunk, humeral motion relative to the scapula, and/or humeral motion relative to the trunk. Problems associated with describing the shoulder position using mathematical terms arise because of the discrepancy in the number of mathematical axes (3) and the number of anatomical descriptors (4), and occasionally because of the interaction between mathematical limitations and the mobility of the shoulder. Several approaches toward measuring shoulder position have been proposed including Joint Coordinate Angles (Grood & Suntay), Euler or Cardan Angles, Helical Axis Decomposition (Woltring), Spherical Coordinates (Cheng), Quaternions, Angle-axis, Rodriguez vectors, and Rotation Matrices. Some of these (Joint Coordinate Angles and Spherical Coordinates) correspond to special cases of Euler Angles, some (Quaternions, Angle-axis, Rodriguez vectors, and Rotation Matrices) are difficult to translate to meaningful anatomical descriptors, and others (Helical Axis Decomposition) work well only within limited ranges of motion. Euler angles function well for many common shoulder movement patterns, but the rotation order that best describes the motion is dependent on the movement pattern. No single solution exists that works well for all shoulder motions. Investigators describing shoulder motion are advised to select and report the rotation sequence and coordinate system orientations that they used to analyze their data.
SESSION 4:

3D Vertebral Kinematics
3D Vertebral Kinematics: Focus on the Equine Back

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Abstract
The back is the centrepiece of the equine musculoskeletal system and hence of crucial importance for performance. There is an increase in reported back problems, but the difficult accessibility of the equine back make diagnosis, therapeutic interventions and the evaluation of the effect of any such interventions into a somewhat hazardous and very subjective procedure. Therefore, (alleged) ailments of the horse’s back are often challenging and unrewarding for the equine practitioner, which makes them at the same time into a relatively easy target for all kinds of alternative therapies with a varying degree of magic by lay people.

A thorough and rather detailed knowledge of the structure of the equine back and a profound understanding of its function are prerequisites to deal successfully with these problems. In this paper an overview is given of the anatomy and physiology of the structures that make up the equine back. Emphasis is laid on dynamical and functional aspects, and on the role of the back within the larger framework of the entire musculoskeletal system. The “bow and string” biomechanical concept of the functioning of the back of quadruped animals is discussed, including its consequences for the use of the horse.

Scientific interest for the biomechanics of the equine back has always been limited, probably not in the least because of the inaccessibility and complexity of the structures that make up the back. Like almost all locomotion research in the horse, scientific work on the equine back came to a halt after World War II. Whereas work on limb kinematics and kinetics was resumed in the early 1970s, the first projects on the back were not undertaken until the late 1980s. Modern technology provides better possibilities for the study of back motion, and in the last decade several groups have published much interesting work. This paper concludes with an overview over the most important recent literature.
Introduction
Interest in equine back problems has increased considerably in recent years. However, they are by no means a new plague to the horse. In 1876 Lupton remarked that back injuries “are among the most common and least understood of equine affections”. Back problems are little understood mainly because of our limitations in the correct diagnosis of these ailments and in the correct interpretation of the abnormalities that are found. The overall vagueness and uncertainties surrounding back problems make them into a difficult and often unrewarding challenge for the equine practitioner. At the same time these factors make back problems into an ideal target for the many lay people with varying degree of trustworthiness, who always surround horse owners telling them that their remedies can provide what regular veterinary practice can not.

To better understand the disorders of the equine back it is imperative to have a profound knowledge of the structures that make up this extremely complex entity. It is perhaps even more important to develop a good insight in the functionality and biomechanical behaviour of what is, literally, the backbone of the equine musculoskeletal system and hence one of the most vital structures for locomotion. Only when such fundamental knowledge exists it will be possible to evaluate objectively various kinds of (alternative) treatments that are advocated as being beneficial without, in most cases, any scientific evidence. Recently, for instance, a case was reported in which manual manipulative treatment was found to have a significant and lasting influence on spinal motion, although the clinical relevance still remained unclear (Faber et al. 2003).

This paper tries to provide some basic information on equine thoracolumbar structure and function. Emphasis will be laid on functionality rather than on anatomical details and more on dynamic than on static aspects of the thoracolumbar structures, i.e. spinal kinematics and the biomechanical concept of the role of the back in locomotion.

Structural components of the back and neck

Bony structures
The spinal column of the horse starts at the occiput of the skull and ends approximately 50 elements later at the end of the tail. Like virtually all mammals, the horse has 7 cervical vertebrae (C1-C7). The atlas (C1) connects the vertebral column with the head and articulates with the epistropheus (C2). These two cervical vertebrae are entirely different in shape compared to the following 5 vertebrae, providing the opportunity for the specific flexion-extension and rotational motions of the head. These motions can relatively easily be checked clinically by manual manipulation after sedation of the horse. After the first two cervical vertebrae, the following vertebrae communicate with each other through the facet joints, each vertebra having a cranial and a caudal set of articular surfaces (fig. 1). In some breeds that are susceptible for the developmental orthopaedic disease osteochondrosis (OC) it has been noted that many OC-lesions can be found in these facet joints. In virtually all cases these lesions are located
at the caudal articular surface (fig. 2) with not more than some kissing-type lesions at the opposing joint surface (the cranial facet of the next vertebra). The clinical relevance of this finding is not known so far.

The 18 (range 17-19) thoracic vertebrae have a much shorter body than the cervical vertebrae. Whereas the latter have no spinous processes, the spinous processes of the thoracic vertebrae increase rapidly in size with Th2-Th9 forming the basis of the withers. Normally, the tip of Th5 or Th6 forms the highest point of the withers. It is of some importance to note that the tips of Th2-Th7 have separate ossification centres that do not close until at a relatively advanced age (>10 years). Occasionally, these ossification centres are mistaken for fractures. The spinous processes have a backward inclination from Th1-Th14. The 15th thoracic vertebra has an upright spinous process, and the processes of the remaining thoracic vertebrae and of the lumbar vertebrae have a forward inclination. Although of little clinical relevance, this change in position is of great functional importance and the observation of this peculiarity has helped in forming the current concept how the back functions (see later). The thoracic vertebrae have many articulating surfaces. Apart from facet joints between the consecutive vertebrae, each rib articulates with two adjacent vertebrae (fig. 3).

The average number of lumbar vertebrae is 6 (range 5-7). Their vertebral bodies are longer than those of the thoracic vertebrae, and their spinous processes shorter. The large transverse processes are characteristic. These are positioned very close to each other and the processes of L5 and L6 normally articulate, as do the processes of L6 with the wing of the ileum. The large transversal processes give this part of the spine little mobility. Further, it is not uncommon that also transverse processes located more cranially articulate, or that articulating processes are fused by bony ankylosis. Although this must influence spinal mobility, it will certainly not always result in a clinical problem.

The sacrum is made up of 5 sacral vertebrae that ankylose during the first 4 or 5 years of life, resulting in a completely rigid structure. The sacrum is connected to the ileal wings by the sacro-iliac joint. This joint is formed by ligamentous structures, which leaves little motion possible, however, the joint is not completely rigid in the normal situation. (Sub-)luxations of the sacro-iliac joint are frequent and often result in asymmetries, clinically known as “hunter bumps”. Although sometimes very evident, they often do not lead to clinical complaints once the situation has stabilised.

The number of coccygeal vertebrae is very variable (15-21). Unlike in other species such as the cat, the biomechanical role of the tail in horses is limited. In the horse, the tail is not known to influence biomechanical behaviour of the back and/or locomotion in a significant way.

Ligamentous structures
There are a large variety of ligamentous structures in the equine back. As passive more or less elastic structures they are essential for the biomechanical role of the back (see later). It can be presumed that, as in the limbs, also in the back the role of ligamentous structures and their interfaces with the bony elements of the musculoskeletal system in
pathologic conditions has been severely underexposed so far. Progression in this area has long been hampered by the difficulties encountered in visualising these structures. It has been shown, however, that many soft tissue abnormalities can be found although the clinical relevance of these has not been firmly established yet in all cases (Denoix 1999). The ligamentous structures that form part of the back can be divided into short and long ligaments.

The interarcual ligaments between the vertebral bodies belong to the short ligaments (fig. 4). They form a kind of elastic interface between adjacent vertebrae and are characterised by the absence of a gel-like nucleus as is common in other species such as dogs and man. Instead, the structure in the horse is fibrous, making classical herniation of the nucleus with ensuing pressure on the spinal cord impossible in this species. The short ligaments also comprise the interspinal ligaments that connect the spinous processes and the intertransverse ligaments between the transverse processes of the lumbar vertebrae (fig. 4). The interspinal ligaments of the horse are non-elastic, except for those in the area of the withers where the spinous processes will make large excursions with respect to each other because of their great length.

The longitudinal dorsal ligament is one of the long ligaments. It runs over the bottom of the spinal canal, i.e. dorsal to the vertebral bodies, from C2 to the sacrum. The ventral longitudinal ligament runs ventral to the vertebral bodies from Th8/9 until the sacrum. Most well-known of the long ligaments is the nuchodorsal ligament that consists of a heavy, elastic structure running from the occiput until the first sacral vertebrae. From the withers it is called the supraspinal ligament. Running over the tips of the spinous processes, it is broad until Th12 and then becomes smaller. A synovial structure between this ligament and the spinous process of Th2, the bursa cucullaris, is often present (fig. 5). Cranial and caudal nuchal bursae may be present too, but are acquired structures (fig. 5). The nuchal ligament is connected to the cervical vertebrae by elastic, laminar structures. The structure as a whole keeps the head in position and is as such an important element in the functional entity that is formed by head, neck and spinal column (see “the back as a functional unit”).

**Muscles**

The great majority of the muscles that attach to bony elements of neck or back run from one part of the axial skeleton to the other, not to parts of the appendicular skeleton. This makes that their primary role is the active, internal stabilisation of the axis of the musculoskeletal system, complimentary to the passive ligamentous structures described above. As a consequence, the system will try to compensate abnormal, asymmetric loading of it (for instance caused by lameness or an unskilful rider) by increased muscle tension. This reaction mechanism of the system explains why, both in man and in the equine species, painful muscle spasms are among the most common and early clinical signs of back problems. It should be realised, however, that such spasms will be in most cases secondary in nature.
There is a deep and a superficial muscle layer. The *M. multifidus* forms part of the deep layer and has been designated as the longest muscle of the body. It consists, in fact, of a long series of muscle bundles, reinforced with ligamentous strips, that run from the sacrum, the lateral sides of the spinous processes of the lumbar vertebrae and the transverse processes of the thoracic vertebrae to the spinous processes of the vertebrae that are located 2 to 6 positions further cranial. The *M. spinalis dorsi et cervicis* is another part of the deep layer running from the spinous processes of the lumbar and the last thoracic vertebrae to the spinous processes of the first thoracic and last cervical vertebrae, thus in the same craniodorsal direction as the *M. multifidus*. The superficial layer is for the largest part formed by the *M. longissimus dorsi*, in volume one of the largest muscles of the body. The muscle originates from the spinous processes of the lumbar vertebrae and runs to the ribs and first cervical vertebrae. In the lumber region there is a union with the aponeurosis of the medial gluteal muscle, which is one of the most important retractors of the hind limb and the most important determinant of the driving force of the hindquarters. A smaller muscle is the *M. iliocostalis dorsi* that runs from the transverse processes of the lumbar vertebrae and the ribs to attach some segments further on the caudal edges of ribs 1-15. The fibre direction of this muscle is cranioventral, so opposing the deep layer.

It should be noted that almost all epaxial musculature is located dorsal to the spinal column. Only the relatively small psoas muscles run under the spinal column from the pelvis to the ventral side of the lumbar and last thoracic vertebrae. There is no musculature ventral to the cranial part of the thoracic vertebral column.

**Innervation**

There are a great many nociceptive receptors located in fascial structures, periosteum and in the numerous ligamentous structures and capsules of articulations. In man it is well known that back pain is as much a pain problem as a back problem, but it is not known whether in horses also other factors than clear-cut pathologies may influence pain perception. No detailed description of the innervation of the structures of the back will be given, but it is important to note that the innervation pattern is segmental in nature. This segmental pattern and the relationship of the various segments system with different organs through the parasympathetic nerve system form the basis of osteopathic practice.

**The back as a functional unit**

The question of how the function of the back can best be described dates back to Antiquity. Galenus (AD 130-200) developed the concept of the “vaulted roof” in which the back and the upper part of the rib cage form a roof over the abdominal and thoracic cavities. A collapse of this roof would be prevented by the spinous processes. However, the fact that the spinous processes do not make contact in the normal situation makes this representation improbable. The next concept was developed by Bergmann in 1847 and further elaborated by Zschokke (1892). This concept implies the representation of the back by a bridge that is resting on 4 piers (the limbs)
The upper ledger represents the supraspinal ligament and withstands tensional forces, the lower one the vertebral bodies, is loaded under compression. The smaller girders between both ledgers represent the spinous processes and the ligaments in between these. Although this concept was generally adhered to until World War II, it contains a basic error in that such a bridge will not be loaded by tension dorsally and by compression ventrally, but just the other way round.

It was the zoologist Slijper who in 1946, after a meticulous study of the anatomical form of the vertebrae and especially of the inclination of the spinous processes in various species, came up with the model that is still holding today. His so-called bow-and-string concept does not only take into account the vertebral column and the limbs, but also the sternum and musculature of the ventral abdomen. In this concept the vertebral column is a bow that is held under intrinsic tension by the ventral abdominal wall (fig. 7). Now the bony vertebral column is loaded under compression and the supraspinal ligament under tension (which is the only force it can resist). Various factors determine the ultimate loading of the system (fig. 8).

Contraction of the abdominal musculature, especially of the rectus abdominus muscle, will tense the bow (i.e. flex the back). Indirectly, the same will be achieved indirectly by the retraction of the forelimbs or the protraction of the hind limbs. The fusion of the aponeuroses of the gluteus medius muscle and the longissimus dorsi muscle as alluded to earlier may be of importance for the latter mechanism. The string will be tensed (i.e. the back extended) by protraction of the forelimbs and retraction of the hind limbs, but also by the considerable weight of the abdominal organs. The latter effect is nicely illustrated by the appearance of many old brood mares with their often very hollow backs. Although in the equestrian circuit many still advocate the contrary, contraction of the epaxial musculature will have an extending effect on the back as well, that is make the back hollower. Given the fact that the vast majority of the epaxial musculature is located dorsal to the thoracolumbar vertebral column, contraction of this musculature will automatically lead to extension of the back.

Of special interest in this biomechanical concept how the back functions is the influence of head and neck movement. If the head is lowered, the nuchodorsal ligament will exert traction on the withers and flex the spinal column (fig. 9a). Vice versa, lifting of the head will extend the back (fig. 9b). Understanding of this concept is of great importance in the athletic training of dressage horses.

**Mobility of the equine spine**

The first modern work on equine gait dates from the late 18th century (Goiffon and Vincent 1779), and, after the breakthrough that was accomplished by Muybridge and Marey more than a century later, many studies have focused on equine locomotion in various aspects. However, studies on equine back movement have been extremely rare until very recently. This is due to the extreme difficulties that are encountered in visualising back movement, together with the relatively small movements of the equine spine, necessitating a high accuracy of the measuring method.
The basic movements of the equine spine are ventro- and dorsiflexion (or extension and flexion in the sagittal plane), lateroflexion or lateral bending, and axial rotation (fig. 10a-c). These movements can be translated at the level of the individual vertebra into 3 rotations in an orthogonal coordinate system: one around the X-axis (flexion-extension), one around the Z-axis (lateroflexion), and one around the Y-axis (axial rotation) (fig. 11).

In the early 80s the biomechanical research group of Doug Leach in Saskatoon published some in vitro work on movement patterns of the equine spine (Townsend et al. 1983, Townsend and Leach 1984). They found that the lumbar part of the spinal column was very rigid, especially with respect to lateroflexion, with increasing mobility in cranial direction. The same applied to axial rotation. It should be emphasised, however, that, although this work is of great value as it gives insight in the potential for movement of various parts of the equine spine, it does not represent reality, as there was obviously no influence of any of the active structures.

Work that was done during the last decade by the groups working at Alfort, in Vienna and in Utrecht has shed more light on the kinematics of the equine back. The first approaches were non-invasive (Licka and Peham 1998, Pourcelot et al. 1998, Audigié et al. 1999). However, the complex and coupled movements of the vertebrae can never be adequately studied using surface markers. For instance, when a horse bends to the left or the right, the spine and hence the individual vertebrae will not only lateroflex, but also be subjected to axial rotation (fig. 12). This complex movement is not deducible from the change in position of a skin marker. In the Utrecht Equine Biomechanics Lab a large invasive experiment was carried out in which horses were first measured using skin markers and then, after placing Steinmann pins in the spinous processes of a number of vertebrae, using markers that were directly connected to the underlying bone. Data were acquired at walk (Faber et al. 2000), trot (Faber et al. 2001a) and canter (Faber et al. 2001b). Based on this data it was possible to develop a method by which real spinal kinematics can be predicted as good as possible from kinematic data obtained using skin markers (Faber et al. 2001c). The technique appeared to be highly reproducible, also when used in different lab settings (Faber et al. 2002).

Spinal motion is by far less at trot than in the other two gaits. At the walk, the range of motion for flexion-extension is fairly constant for vertebrae caudal to Th10 (approximately 7º), lateral bending is most outspoken in the cranial thoracic vertebrae and in the pelvic segments (values up to 5.6º), but less in the lumbar region between Th17 and L5 (<4º). Axial rotation increases gradually from 4º at Th6 to 13º at the tuber coxae. At the trot the range for flexion-extension for all vertebrae does not pass 2.8-4.9º, lateral bending is even less (1.9-3.6º). Axial rotation at this gait is in the order of 3º. At the canter flexion-extension movement is substantially larger (maximal range 15.8±1.3º). Lateral bending is maximally 5.2±0.7º and axial rotation 7.8±1.2º.

It should be noted that the shape of the curves partly depends on the gait. Flexion-extension movement has a double sinusoidal motion pattern at the walk and trot, but a single sinusoidal pattern at the canter. Lateroflexion has
the form of a single peak and trough at all gaits, as has axial rotation. This has to do with the symmetry of the gait and the effect of hind limb placement on spinal kinematics. Typical examples of spinal motion patterns are given in fig. 13.

Variability within the same horse is limited for flexion-extension and axial rotation (6-8%), but considerably more for lateroflexion (8-18%). The variability between horses is larger, as could be expected, and the same applies here: lateroflexion may vary as much as 16-25% between individual horses, which is considerably more than the variation in the rotation around the other two axes (10-16%) (fig. 14).

Recent developments
Interest in function and dysfunction of the equine back has been growing rapidly from the second half of the 1990s onwards, resulting in an increasing number of publications on the topic. The increase in scientific output is driven by the growing awareness of the importance of the back for equine locomotion and performance, being the connecting element of the limbs, and has been made possible by the rapid technological developments, including an enormous increase in computational capacity, in recent years. The work that has been done and is currently being carried out comprises various areas of back physiology and pathology, covering topics ranging from in vitro work and the development of field techniques to the creation of mathematical models of the equine spine.

In vitro research
Acknowledging the role of anatomical constraints as the basis for all motions of the equine back, Degeuerce et al. (2004) have continued the line of in vitro research initiated by Townsend et al. (1983), and further pursued by Denoix (1987, 1992, 1999), in studying the motion of the sacro-iliac joint (SIJ) in anatomical specimens. The SIJ is well known to play an important role in human lower back pain (Pool-Goudzwaard et al. 2003), but in the horse, although the SIJ is often incriminated as a cause for back pain, even its range of motion had never been established. It resulted that there is only a very limited amount (on average less than one degree) of nutation (which is the rotation of the pelvis in relation to the sacrum around the X-axis (perpendicular to the sagittal plane) of the latter bone), related to the much larger dorsal-ventral flexion motion (approximately 24°) of the lumbosacral joint (LSJ). When the sacrosciatic and sacrotuberal ligaments are severed, this range of motion almost doubles. It is concluded that SIJ motion is much less than expected based on earlier work (Faber et al. 1999), probably because of a different analysis technique, and that these small amounts of motion most probably are too small to be measured in vivo.

The relationship of back problems with lameness
Back problems are in the clinical setting often related to poor performance and sub-clinical lameness, especially of the hindquarters. Although there is common agreement that there is a mutual influence of lameness on back function and vice versa, the extent of this relationship and the mechanistic aspects of it are far from clear. In a recent survey a group of orthopaedic
patients and a control group of animals presented for a pre-purchase exam (as the most appropriate control population) were subjected to a full lameness examination as well as a full back examination, irrespective of their eventual complaints. In that study, which was one of the first field studies on the subject, it appeared that the prevalence of lameness in horses with a diagnosed back problem was much higher than in horses without such a problem. Of 805 horses presented as suspected of an orthopaedic ailment, in 208 (26%) there were indications for both lameness and back pain (Landman et al., 2004). These figures do not give, however, evidence about a causal relationship. In another recent study Dyson (2005) diagnosed concurrent forelimb and or hind limb lameness due to an unrelated cause in 23 (46%) of horses with primary thoracolumbar or sacroiliac region pain.

Causal relationships between lameness and back movements can only be demonstrated in experimental studies. It has been shown that experimentally induced forelimb or hind limb lameness may alter the biomechanics of the equine back (Buchner et al. 1996, Pourcelot et al. 1998). Very recently it was shown, however, that when very subtle lameness is induced (maximally 2/5), the influence on back motion is very limited (Gómez Alvarez et al., in preparation). This makes sub-clinical lameness as a cause for back pain improbable. It should be stated though that all these studies have investigated the acute effect of induced lameness on back motion whereas in the clinical setting it can be presumed that long-term effects of chronic lameness are more important.

Few studies so far have focused on the effect of induced back pain. Jeffcott et al. (1982) induced reversible back pain in Standardbred trotters by the injection of a high concentration of lactic acid in the epaxial musculature. They did not see effects on linear and temporal stride parameters (stride length, stride frequency, pro- and retraction angles). In a very recent study in Dutch Warmbloods, in which the same technique to induce back pain was used, this observation could be confirmed (Wennerstrand et al., in preparation), but in a study on patients suffering from back pain as diagnosed by repeated palpation a shorter stride length at walk (not at trot) was found in the back pain patients compared to healthy controls (Wennerstrand et al. 2004). In both studies a significant influence of back pain on spinal kinematics could be noted. In the short-term effect as studied by the experimental study there seems to be a difference in the pain response directly after the induction of back pain, and the resulting muscle stiffness occurring during the following days (Wennerstrand et al., in preparation).

**Saddle pressure**
The role of the saddle in riding has been a concern for ages, and (weighted) saddles are known to influence back motion (DeCocq et al. 2004). The introduction on the market of devices that can be used to measure the pressure under the saddle has received considerable attention as this technique was considered to be very helpful in quantifying the very difficult and so far highly subjective area of assessing saddle fit and in studying the relationship of rider and horse (Harman 1994, 1997). Potentially promising,
the technique is still in its infancy in which the potential possibilities of the technique are explored (Werner et al. 2002) and normal pressure patterns established (Fruehwirth et al. 2004). Although the first report (in the square standing horse!) on the validity of this kind of systems was encouraging (Jeffcott et al. 1999), it has become clear that the systems may not be as reliable as hoped for (DeCocq et al. 2005). The practical application of these systems in moving and athletically performing horses is fraught with pitfalls due to insufficient fit of the pad to the shape of the equine back, wrinkling of the pad, shear forces that are unaccounted for but interfere with outcome, size of load cells, maximal measuring capacity of load cells, and other technical factors. There is little doubt that these technical problems will be overcome and one may even think of saddles with in-built sensors in relevant areas in the future, but at present the systems cannot be considered to give unequivocal and reliable results.

The back and performance
The empirically established relationship between conformation (and visually assessed gait, or kinematics in other terms) and performance has formed the basis for the selection of breeding stock ever since man began to purposefully breed horses after domestication (Holmström 2001). Recently, back kinematics has received special attention in relationship to jumping performance. Cassiat et al. (2004) were able to show significant differences in back kinematics between two groups of show jumpers performing at a high and a low level respectively. The good performers appeared to pitch their back line less forward during the forelimb stance phase before take-off and to straighten it more after landing, probably indicating a more efficient strutting action of the forelimbs. Bobbert et al. (2005) reported a different back action between the best and worst performers in a puissance competition. At hind limb clearance, the best jumpers had their centre of gravity further beyond the fence. This was possible because the best jumpers had a greater angle between the trunk and the hind limbs. This greater angle, in turn, was partly attributed to increased flexion in the knee joints, but partly also to a 0.1 rad further tilting of the sacrum. The observations by these studies findings lend support to the current practice of having visual inspections of submaximal free jumps at foal age in the process of selection of talented show jumpers.

A completely different relationship of back kinematics to performance is the effect of specific forms of equestrian activity on the back. In dressage, horses are required to compete in certain positions that do not correspond to the natural position that the horse would assume of its own free will. This is especially obvious with respect to the position of the head and neck, which is described in the rules of the Fédération Equestre Internationale (FEI) as: “The neck should be raised, the poll high and the head slightly in front of the vertical”, implying a much more upright position than in the natural situation. Guidelines for the correct position of the head and neck in dressage have been given and discussed for ages in the equestrian literature (Cavendysh 1674, de Solleyseel 1733, Lenoble du Teil 1889, Decarpentry 1971), but still continue to be a source of controversy at present times (Balkenhol et al. 2003; Janssen 2003). In this discussion the
acceptability from an animal welfare viewpoint of certain training techniques implying head/neck positions strongly diverging from the natural position take a prominent position. In a recent comprehensive study carried out by an international consortium that included kinematic and kinetic parameters and saddle pressure data on the effects of head and neck position, it could be established that there was a significant influence of head/neck position on thoracolumbar kinematics, principally in the sagittal plane. Positions with an elevated neck tended to induce extension in the thoracic region and flexion in the lumbar region. Lower neck positions produced the opposite. High neck positions generally led to a restriction of range of motion (ROM) of vertebrae, especially in the lumbar area, but low-neck positions increased ROM (fig. 15). A very high position of the neck seemed to greatly disturb normal kinematics, much more than a strongly flexed position (Gómez Álvarez et al., accepted). These results confirmed earlier work concerning a more restricted number of head and neck positions (Rhodin et al. 2005). It is anticipated that more work will be done in this important area, the outcome of which may influence the policy of sport-regulating bodies such as the FEI as well as, and interrelated with this, public opinion about the ethical acceptability of certain equestrian activities (van Weeren 2005). The importance of the issue was underlined by the recent (2006) workshop organized by the FEI at their Lausanne headquarters in which the acceptability of the Rollkür" or “over bended” training technique, in which the horse is ridden with a strongly flexed mid-cervical region that brings the head almost down between the front limbs, was discussed. The preliminary outcome was that “there was clearly no evidence at the present time that any structural damage is caused by this training exercise, when used appropriately by expert riders” (Jeffcott 2006).

**Mathematical modelling**

The relative ease with which quantitative data can be collected and the enormous increase in computational capacity since the large-scale introduction of computers about two decades ago, allow for the development of complicated mathematical models of biological structures and their function. Modelling on the basis of finite element analysis is commonplace in engineering, but is more complicated in living tissues, which almost always consist of heterogeneous materials that in some cases may undergo morphological changes due to voluntary movements. Realistic modelling is only possible if input parameters, among which material properties, are correct. Currently, studies are undertaken to determine material properties, such as stiffness, of the equine back (Schlacher et al. 2004). These, and other input parameters can serve as a basis for the modelling of the effects of certain events, for example a rider, on the back (Peham and Schobesberger 2004). Though far from perfect at present, models of the equine back may increase in importance in the future, as their accuracy grows, to evaluate and predict the effects of certain interventions on the function of the equine back.

**Conclusion**
It can be concluded that the equine back is a very complex structure that takes a central position in the entire equine musculoskeletal system and hence can be decisive for performance. A good knowledge of how the back functions, is therefore essential. The close interrelationship between the limbs and the back (and neck) is often underestimated and not yet fully understood. The biomechanical concept of the action of the back is crucial too for a good understanding of various riding techniques in dressage horses and how certain training methods could affect the musculoskeletal system. It should be emphasised, however, that this matter is extremely complex, as there are many factors that may influence performance. Of these, use of the horse, the quality of the rider and the tack are among the most important.

In recent years, much progress in the understanding of the functioning of the equine back has been made using a variety of technical approaches. It can be anticipated that, notwithstanding the limitations all kinematic gait analysis systems suffer (van Weeren 2002), computerised analysis of spinal kinematics will become more popular as an aid in diagnosis and to monitor recovery and/or the success of chosen therapies.

References


Legends for illustrations

**Figure 1.** C5 as a typical example of one of the cervical vertebrae after C2. Note the absence of spinous processes and the large facet joints. 1: cranial articular surface facet joint; 2: caudal articular surface facet joint. (Modified from: Nickel, R. *et al.* (1961) *Lehrbuch der Anatomie der Haustiere.* Band I Bewegungsapparat. 2. Auflage. Berlin, Paul Parey.)

**Figure 2.** Osteochondrotic lesion in the caudal facet of C6 of a 5-month-old Dutch Warmblood foal.

**Figure 3.** Th8 and Th9. Note the large spinous processes that form the basis of the withers. The ribs articulate with two thoracic vertebrae each. 1: articular surface of cranial articulation with rib; 2: articular surface of caudal articulation with rib. (Modified from: Nickel, R. *et al.* (1961) *Lehrbuch der Anatomie der Haustiere.* Band I Bewegungsapparat. 2. Auflage. Berlin, Paul Parey.)

**Figure 4.** Ligamentous structures between vertebrae. Almost all of these structures have a rather limited elasticity. (Modified from: Denoix, J.M. and Pailloux, J.P. (2001) *Physical Therapy and Massage for the Horse.* 2nd Ed. London, Manson Publishing.)

**Figure 5.** Nuchal ligament and connection between the nuchal ligament and the cervical vertebrae. The *bursa cucullaris* (1) is a regular bursa that is mostly present. The cranial and caudal nuchal bursae (2,3) are acquired bursae. (Modified from: Nickel, R. *et al.* (1961) *Lehrbuch der Anatomie der Haustiere.* Band I Bewegungsapparat. 2. Auflage. Berlin, Paul Parey.)

**Figure 6.** The (erroneous) bridge concept of the vertebral column as depicted by Krüger. Open arrows represent tensile forces, closed arrows compressive forces. (From: Krüger, W. (1939) Über die Schwingungen der Wirbelsäule –insbesondere der Wirbelbrücke- des Pferdes während der Bewegung. *Berl. Münchn. Tierärztz. Wschr.* 13, 129-133.)
Figure 7. “Bow and arrow” concept of the back according to Slijper. The vertebral column is the bow and the ventral musculature and sternum are the string. The ribs, lateral abdominal musculature, spinous processes and ligamentous connections are additional elements. (From: Slijper, E.J. (1946) Comparative biologic-anatomical investigations on the vertebral column and spinal musculature of mammals. Proc. K. Ned. Acad. Wetensch. 42, 1-128.)

Figure 8. Factors that determine the motion of the back according to the “bow and string” concept.

Figure 9. The effect of lowering (a) and raising (b) of the head on flexion and extension of the back. (Modified from: Denoix, J.M. and Pailloux, J.P. (2001) Physical Therapy and Massage for the Horse. 2nd Ed. London, Manson Publishing.)

Figure 10. The three basic movements of the equine back: flexion/extension (a), lateroflexion or lateral bending (b), and axial rotation (c).

Figure 11. The basic movements of the back depicted as rotations of an individual vertebra around the three axes of an orthogonal coordinate system.

Figure 12. Bending of the vertebral column cannot be effectuated by lateral bending alone; there is always a concomitant axial rotation.

Figure 13. Mean motion patterns of 3 vertebrae (Th10, L1 and S3) of 5 horses walking on a treadmill at a speed of 1.6 m/s. The stride cycle is represented by the bars below (LH: left hind; RH: right hind; LF: left fore; RF: right fore). The bar is closed when the limb is in contact with the ground (stance phase). A: flexion/extension; B: lateral bending; C: axial rotation.

Figure 14. Intra-individual and inter-individual variation with respect to the three basic movements.

Figure 15. Mean flexion extension (FE) angular motion pattern (AMP) of one horse (T10 at walk). The curves represent head-neck positions 4 (HNP4, a position close to the “Rollkūr position, defined as “Neck lowered and flexed, bridge of the nose considerably behind the vertical”) and 5 (HNP5, the most opposite position defined as “Neck extremely elevated and bridge of the nose considerably in front of the vertical”), and their controls (indicated as HNP1, the free or natural position. Of this position a speed range was made in order to be able to compare every head-neck position with a speed matched control).
Figure 3
Figure 9
Figure 11

Figure 12
Figure 13

Figure 14
Figure 15
3D Kinematics of the Equine Temporomandibular Joint

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McPhail Equine Performance Center, Michigan State University

The objectives of this study were to develop a method of measuring 3D kinematics of the equine TMJ during chewing and to describe the 3D movements of the mandible in terms of a virtual marker created between the mandibular rami to represent motion of the mandibular cheek teeth.

Four mature horses in good dental health were used in the study. In the absence of algorithms correcting for skin displacement in the equine skull and mandible, pilot studies were performed in which markers were placed at sites where skin displacement was expected to be small based on the fact that skin adhered tightly to the underlying bone without the intervention of copious loose soft tissue. Relative displacements between a large number of marker locations were determined, and locations with the least amount of displacement were chosen for our study.

Twelve spherical, reflective, tracking markers were attached to the skin overlying palpable bony landmarks. Two tracking markers were placed on the dorsal midline of the face at the level of the orbit and 10 cm further rostrally. Five tracking markers were placed bilaterally on the skull over the middle of the facial crest and the rostral part of the facial crest. Markers were placed bilaterally on the mandible at the notch where the facial vessels cross the ventral edge of the mandible, and dorsally and ventrally on the caudal edge of the mandibular ramus. In addition, stationary files were recorded with temporary markers located over the right and left articular tubercles of the skull and the right and left condylar processes of the mandible. These stationary files were used to calculate locations of the articular tubercles and condylar processes during chewing from the positions of the tracking markers. Kinematic data were collected at a sampling rate of 120 Hz within a calibrated volume measuring approximately 2 m x 1 m x 1 m. A single trial consisted of a minimum of 4 chewing cycles and 6 trials/horse were recorded.

Orthogonal coordinate systems, based on anatomic planes, were established for the skull and mandible so that the motions could be described in meaningful anatomic terms. For the skull, the y axis was a vector running from the right to the left articular tubercle. The x axis pointed rostrally and was the average of unit vectors established from the caudal marker to the rostral marker on the left and right facial crests. The z axis pointed dorsally and was perpendicular to the x and y axes. The mandibular BCS was parallel to the BCS for the skull, with the y axis being a vector from the right to left mandibular condyle.

Movements of the mandible relative to the skull were described in terms of 3 successive rotations: pitch, roll, and yaw, where pitch is a rotation about the
transverse, horizontal (y) axis, roll is a rotation about the longitudinal (x) axis, and yaw is a rotation about the vertical (z) axis.

A virtual midline-mandible marker was created and tracked relative to the skull’s coordinate system to aid in viewing mandibular motion and to facilitate comparisons with previous studies in which relative motion between skull and mandible was estimated by tracking markers on the upper and lower lips. This virtual marker was located on the midline between the rami of the mandibles at the level of the rostral end of the facial crest and at the height of the marker on the mandibular notch.

Data were processed for the first 3 complete, consecutive chewing cycles in each trial. A cycle was defined from 1 minimum pitch angle to the next, which coincided with the mouth being fully closed. As the mouth opened, the pitch angle increased and, as the mouth closed again, the pitch angle decreased and returned to a minimum value. The chewing side was determined as the side toward which the mandible moved during the closing stroke. Data were normalized by subtracting the output from the stationary files from the output from the tracking file for all trials. In accordance with the usual convention, the chewing cycle was divided into 3 phases or strokes. The opening stroke occupied the time between the minimum and maximum pitch angles (mouth fully closed to mouth fully open). It was followed by the closing stroke, which ended when the yaw angle reached its largest absolute value which represents maximal lateral displacement of the mandible (positive for chewing on the left, negative for chewing on the right). The power stroke, during which the occlusal surface of the mandibular arcade grinds across the occlusal surface of the maxillary arcade, occupied the time between the end of the closing stroke and the start of the next opening stroke.

During data collections, two horses chewed only on the left and two chewed only on the right. A chewing cycle began when the pitch angle reached a minimum, which corresponds with the jaw in its fully closed position. For a horse chewing on the right, during the opening stroke, the mandible rotated clockwise as seen from the right (positive pitch) until a maximum pitch angle was reached, which corresponds with the fully open position. The direction of mandibular rotation was reversed (negative pitch) during the closing stroke. A change in slope of the pitch angle as the value decreased represented the transition between the closing stroke and the power stroke. When the pitch angle reached a minimum value, the mouth was closed again and the first cycle was complete.

During the opening stroke, the virtual midline mandibular marker moved ventrally, laterally toward the chewing side, and slightly caudally. During the closing stroke, the marker moved dorsally, medially, and slightly rostrally. During the power stroke, the mandible slid medially and dorsally as the mandibular cheek teeth moved across the occlusal surface of the maxillary cheek teeth. The 4 horses had similar chewing patterns but the amplitudes varied among horses (tables 1 and 2).
The results indicate that the downward hinge movement of the mandible (a pitch rotation about the transverse horizontal axis) during the opening phase was accompanied by small rotations in the other 2 axes. During the opening phase, the rolling motion seen from the rostral view, separated the upper and lower dental arcades on the chewing side. The yaw motion swiveled the mandible (crossed the jaw) away from the chewing side. During the closing phase, a small amount of roll assisted in bringing the upper and lower arcades into apposition on the chewing side. The yaw swiveled the mandible toward the midline through the closing and power strokes, which slid the lower arcade across the upper arcade in a lateral to medial direction. The slope of the molar table angle likely dictates the extent of roll angle during the power stroke; a steeper molar table angle will result in more mandibular roll and a more level molar table will have less mandibular roll.

It is concluded that the equine TMJ allows considerable mobility of the mandible relative to the skull during chewing. The method presented in this report can be used to compare the range of motion of the TMJ among horses with TMJ disease or dental irregularities or within an individual horse before and after dental procedures.

Reference

Table 1  Mean (SD) values* for mandibular rotational and translational ranges of motion in 4 horses.

| Variables                        | Horse 1          | Horse 2          | Horse 3          | Horse 4          | Group mean
|----------------------------------|------------------|------------------|------------------|------------------|-------------
| Ranges of rotational motion      |                  |                  |                  |                  |             |
| Pitch (°)                        | 2.25(0.44)       | 4.58(0.54)       | 3.45(0.08)       | 2.58(0.22)       | 3.22(0.32)  |
| Yaw (°)                          | 2.56(0.78)       | 3.12(0.33)       | 3.02(0.06)       | 3.55(0.14)       | 3.06(0.33)  |
| Roll (°)                         | 0.55(0.10)       | 1.05(0.25)       | 1.26(0.08)       | 0.72(0.15)       | 0.90(0.15)  |
| Ranges of translational motion   |                  |                  |                  |                  |             |
| Rostrocaudal (cm)                | 7.58(0.55)       | 12.87(2.25)      | 12.30(0.51)      | 6.65(0.82)       | 9.85(1.03)  |
| Mediolateral (cm)                | 2.64(0.66)       | 10.55(2.14)      | 4.64(0.64)       | 7.59(1.25)       | 6.36(1.17)  |
| Dorsoventral (cm)                | 2.35(0.71)       | 4.36(0.82)       | 4.21(0.55)       | 2.42(0.21)       | 3.34(0.57)  |

*Values for individual horses are means of 4 trials, each with 3 chewing cycles. Values for group are means of the 4 horses.
Table 2 Mean (SD) values* for displacements of midline-mandibular marker in 4 horses.

<table>
<thead>
<tr>
<th>Displacement</th>
<th>Horse 1</th>
<th>Horse 2</th>
<th>Horse 3</th>
<th>Horse 4</th>
<th>Group mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rostrocaudal (cm)</td>
<td>3.32(0.14)</td>
<td>5.14(0.51)</td>
<td>4.40(0.12)</td>
<td>5.07(0.48)</td>
<td>4.49(0.31)</td>
</tr>
<tr>
<td>Mediolateral (cm)</td>
<td>11.45(0.25)</td>
<td>17.59(0.69)</td>
<td>12.56(0.40)</td>
<td>15.96(1.74)</td>
<td>14.39(0.77)</td>
</tr>
<tr>
<td>Dorsoventral (cm)</td>
<td>12.65(0.18)</td>
<td>17.66(0.67)</td>
<td>12.20(0.28)</td>
<td>11.99(1.24)</td>
<td>13.62(0.60)</td>
</tr>
</tbody>
</table>

See Table 1 for key.
SESSION 5:

Solutions to the Skin Displacement Issue (horse and human)
Soft Tissue Artefact Assessment of Marker Clusters on the Lower Forelimb During in vitro Movement

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Reasons for performing the study
Non-invasive assessment of three dimensional (3D) movements of humans executing different tasks is becoming more commonplace in biomechanics, particularly with recent advances in techniques, software and hardware. Progress in 3D non-invasive assessment of equine limbs has been made by some authors, (Clayton et al., 2004; Johnston et al., 2004) with correction algorithms currently available for skin markers on the tibia and third metatarsus, (Lanovas et al., 2004) and the equine radius, (Sha, et al., 2004). Improvements in bone pose estimation are possible, but solutions tend to be marker (location and configuration), task and segment specific. Soft tissue artefacts of lateral markers on the lower forelimb in two dimensions have been assessed, (van Weeren et al., 1988), but correction algorithms are not available for marker clusters or for 3D movements.

Hypothesis or objectives
To quantify 3D soft tissue artefacts on segments of the lower forelimb from marker clusters fixed to rigid plates and attached to the skin.

Methods
Seven normal forelimbs removed at the articular surface of the third metacarpal bone (McIII) were collected from horses subjected to euthanasia for reasons unconnected to the study. Bone screws were fixed to McIII, the proximal phalanx (PI) and middle phalanx (PII) and the superficial digital flexor tendon (SDFT) and deep digital flexor tendon (DDFT) were tightened against McIII proximally using a jubilee clip. The isolated limbs were secured in cadaver rig described by Hobbs et al., (2004), based on an Instron model 8502 (Instron, UK).

Six infra-red cameras (ProReflex®, Qualysis Medical AB, Goteburg, Sweden) were positioned in a semi circle around the Instron and calibrated. Four segments were defined for the limb: hoof (PIII), PII, PI and McIII. For PII, PI and McIII segments, a rigid marker cluster with four retro-reflective markers was inserted into the bone screw and a rigid marker cluster with four retro-reflective markers was attached to the skin. A static trial was recorded with additional anatomical markers on the medial and lateral joint lines. These anatomical markers were removed, except for the hoof and kinematic data were then recorded at 120 Hz from movement in impact, midstance, midstance with 10° of heel lift, midstance with 5° of medial-lateral imbalance and breakover positions.
The completed trials were then digitised using QTM (Qualysis Medical AB, Goteburg, Sweden) and exported as a C3D file to Visual 3D (C-Motion Inc., Gaithersburg, USA). The kinematic data was smoothed with a Butterworth 4th order filter with a cut off frequency of 10 Hz. For each limb an ensemble average was computed from ten replicates of the movement. Marker clusters were interrogated and segment kinematics with respect to the lab coordinate system (LCS) were calculated for segments using the Calibrated Anatomical System Technique (CAST), Cappozzo et al. (1995) where photogrammetric errors were within acceptable limits (max of ≤1mm). 3D segment rotations of the bone marker clusters compared to the skin marker clusters were plotted and the differences in magnitude and shape agreement were recorded.

**Results**
Larger bone rotations were recorded for PII about the flexion-extension axis with fair shape agreement for all positions with values of up to -7.37 (3.34)° difference (mean (SD)). Excellent and fair to excellent shape agreement was found about the flexion-extension axis with mean values of below 2° and 0.4° for PI and McIII respectively. In abduction-adduction bone rotations were mainly larger -0.41 (0.70)° with -0.47 (0.49)° for PII and McIII, but with fair to poor shape agreement whereas better shape agreement was found for PI, but with larger skin rotations. More skin rotation and poor shape agreement was generally found in axial rotation for PII and PI with the largest values found for PI 3.78 (3.26)°. Good to fair shape agreement was found for McIII with up to 1.57 (1.36)° difference, except at impact where shape agreement was poor.

**Conclusions and potential relevance**
These results suggest that skin marker clusters closely represent the movement of the underlying bone for the MPJ in flexion-extension, but may result in overestimation of the PIPJ and underestimation of the DIPJ. Improvements to the geometry and inertial properties of PI marker cluster to reduce soft tissue artefacts may provide a method of estimating abduction-adduction, but as rotations of the MPJ and PIPJ are relatively small (Chateau et al., 2004a; 2004b) errors may be of the same order of magnitude as the measurements. Consequently, further investigation of soft tissue artefacts is necessary with improved skin marker clusters to assess the accuracy of reporting rotations in abduction-adduction. At present axial rotations of PII and PI segments cannot be reliably estimated using this method, but an indication of McIII rotation may be possible. Refinements to the technique are required to improve data reliability and further work will include an investigation of correction methods.

**Reference:**


Development and Verification of Correction Algorithms for Skin-based Markers for 3D Kinematic Analysis of the Equine Tarsal Joint

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Skin based markers are preferred to bone-fixed markers in kinematic studies but the errors due to skin displacement relative to the underlying bones can create serious errors in the data. This goal of this study was to investigate the feasibility of using skin-based markers for 3D analysis of tarsal joint motion.

Triads of reflective markers were attached to pins that were rigidly fixed to the tibial and MT3 segments of three horses. In addition, six retroreflective markers were attached to the skin over each of the tibial and metatarsal segments in three horses. For each segment, a cluster of 3 markers was placed over the proximal part of the segment and a cluster of three markers was placed over the distal part of the segment. The bone-fixed markers and the skin markers were tracked automatically as the horses trotted along a rubberized runway. The 3D skin displacement patterns were parameterized using a truncated Fourier series model with the displacements being expressed in terms of the local coordinate system for each bone. To minimize the effects of any marker placement variation between subjects, the standing pose value of the coordinate (expressed as % segment length) was subtracted from the skin displacement prior to fitting the data to the Fourier series.

Skin displacement artifacts were observed in all three axes of each segment, with the largest displacements occurring at the proximal tibia. Mean skin displacement amplitudes in the tibia were 6.7%, 3.2% and 10.5% of segment length, and for MT3 were 2.6%, 1.4% and 3.8% of segment length for the craniocaudal, mediolateral and longitudinal axes, respectively. Displacements were consistent between subjects, which should allow them to be used as a basis for developing a correction procedure for 3D tarsal kinematics.

A second set of data was collected at a later date with the horses moving at the same gait and speed and with skin markers in the same locations as for the previous study. 3D motions of the markers were corrected for skin displacement using the algorithms developed previously. Marker locations that were uncorrected for skin displacement (U) were compared with marker locations after correction for skin displacement (C) and with previous data from bone-fixed markers (B). The two sets of curves were compared separately during stance and swing phases by correlation, root mean square errors (RMS) and shape agreement.

The RMS of B and C were smaller than those of B and U for all motions. The correlation coefficients of B and C were higher than those of B and U. Shape
agreement was good for flexion/extension in stance and swing and for abduction/adduction in swing, fair for adduction/abduction in stance and poor for internal/external rotation in stance and swing.

It was concluded that, with appropriate correction for skin movement relative to skeletal landmarks, the skin marker set used in this study can identify tarsal 3D motions for flexion/extension and abduction/adduction but not for axial rotation.

References


History and 2D Solutions in Equine Kinematic Studies

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Introduction

The very first evidence of the awakening artistic potential of the species *Homo sapiens* shows man’s fascination with the animal species surrounding him. Virtually all cave paintings and rock art that has been discovered thus far, and that may date back as many as 30,000 years, depict scenes of animal life. The first evidence of some scientific interest in the way animals move can be found in Aristotle’s *De incessu animalium*, in which he describes the footfall pattern of a walking quadruped (Aristotle 384-322 BC).

Some works from the late Middle Ages or the 17th or 18th century mention a few lines on gait analysis and there is one remarkable book that is in its entirety dedicated to the subject (Goiffon and Vincent 1779), but the real work does not start before the invention of photography in the middle of the 19th century. Of great importance was the encounter of photographer Eadweard Muybridge and railroad magnate Leland Stanford. Stanford had made his money on the quickly expanding railway network of the U.S. as owner of the Union Pacific Railroad, but he was a lover of horses as well. He owned a famous and successful trotter, named *Occident*, and was intrigued by the question whether or not there was an airborne phase in trot. It is known that this item had already been raised by the old Egyptians (Simpson 1951), and only a few years earlier it had been the topic of a heated dispute between Joseph Gamgee from Edinburgh and Neville Goodman from Glasgow (Gamgee 1869, 1870; Goodman 1870, 1871), but the temporal resolution of the human eye was simply not high enough to solve the problem. Stanford asked Muybridge to solve the problem for him, using photography. He finally succeeded in doing so by using a linear array of 24 cameras that were triggered by the horses themselves when they passed by and by reducing shutter speed to an incredible 1/6000 second (in a time when an exposure of ½ second was considered instantaneous!) (Figs.1,2).

After the invention of cine film a few decades later, which was largely based on the work of the other great pioneer of gait analysis, Etienne Marey from France (Marey 1882), research in the area increased, especially in pre World War II Germany (Schmaltz 1922, Walter 1925, Krüger 1937, 1938). After the first merely descriptive work, research was gaining more depth, but, as a consequence, then ran into the problem of skin displacement.

Skin displacement

*History*
All photographic and cinematographic techniques relied on markers that were somehow fixed or applied to the skin overlying certain bony landmarks. However, it was already noted in 1910 by Fick that this procedure introduced a certain error. Fick stated that: “In the first place it is difficult to mark on the moving limbs certain sites that do not alter their positions on the body surface during movement, as the skin near the joints shifts considerably” (Fick 1910).

The problem was recognized and taken seriously, but was not easy to solve. Skin displacement is a cyclic movement with the same frequency as the step cycle and can therefore not afterwards be detected in the raw data (Capozzo et al. 1988). Walter (1925) even limited his studies to the distal parts of the limbs because of the large amount of skin displacement in the more proximal parts. Krüger (1937, 1938) used the original approach of oiling the skin and using oblique lighting for his cinematographic studies. In this way he could visualize the underlying bone through the creation of sharp contrast between the protruding parts of the bone and the surrounding skin. However, the technique only worked in certain areas and he was condemned to the use of very skinny horses.

After World War II the interest in equine gait analysis declined together with the horse’s significance in its traditional fields of employment: warfare, transport and agriculture. However, from the mid 1960s onwards the horse made a glorious comeback as a sports and leisure animal end interest in the analysis of gait revived, now greatly aided by the huge advances in computing sciences (van Weeren 2001). The skin displacement artifact was again recognized (Fredricson et al. 1972), and a first in-depth study of the problem was started at Utrecht University (The Netherlands) in the mid 1980s.

Two-dimensional analysis of skin displacement – the distal limbs
For the quantification of the skin displacement artifact in the distal parts of the extremities of the horse an original surgical approach was chosen. Light-emitting diodes (LEDs), fitted in stainless steel tubs, were implanted in the bones underlying the skin at the places of interest. The lights of these LEDs shone through the skin, which itself was marked with conventional markers, allowing the simultaneous visualization of the position of both skin and bone (Figs. 3a-c). Data were captured by way of a motor-driven photo camera, creating series of photographs not unlike those produced by Muybridge more than a century earlier (van Weeren and Barneveld 1986). The relationships between relative time (T), joint angle (A₁…Aₙ), and one or more 2-dimensional skin displacement (X₁ Y₁….Xₙ Yₙ) were sought using regression analysis on a series of photographs from the same experiment. Displacement could be calculated as:

\[
\text{Displacement (X or Y) = p₀ + p₁A₁ + p₂A₂ + \ldots + pₙAₙ.}
\]

As more angles did not bring additional value, a simple regression model:
Displacement $p_0 X$ or $Y(t) = p_0 + p_1 A(t)$

could be used. Because $p_0$ only gives information about the initial position of the LED and average displacement is zero, the equation became finally:

$X(t) = p_1 \langle A(t) - \langle A(t) \rangle \rangle$, with $\langle A(t) \rangle$ the average of joint angle $A(t)$ during a stride.

It appeared that all displacements along the Y-axis (perpendicular to the long axis of the bone) were less than the detection level of 2mm, except for the distal tibia (total displacement 9.7mm). Displacements along the X-axis varied from 2.6mm (phalanx I hind) to 20.6mm at the distal tibia. Other sites with displacements $> 10$mm were the distal radius and the proximal metatarsus (van Weeren et al. 1988).

Two-dimensional analysis of skin displacement – the proximal limbs

The technique used in the distal parts of the limbs could not be used in the more proximally located parts of the limbs because it required a surgical approach from the medial side of the limb (so as not to interfere with skin displacement laterally), which was not possible at the proximal ends of the limbs. Skin displacement was much larger, and thus more interesting, however, in the proximal parts of the limbs. For this reason, an entirely different approach was used to quantify skin displacement in those areas. This technique used a Steinmann pin that was implanted in the bone at a site far away from the areas of interest where skin displacement had to be measured. To this pin a rigid plastic strip with two reference markers was attached, thus creating a rigid body consisting of bone, pin and strip with reference markers. The position of those reference markers therefore was an indirect indication of the position of the bone to which the position of skin markers could be related (Figs. 4a,b). For the analysis, the displacement of the skin markers relative to the plastic strip measured on serial photographs was determined, thereby taking the average position with respect to the strip during a stride as “zero displacement”. Displacements were expressed as X, Y positions in a bone-related coordinate system with the bone axis as X-axis. Temporal data were provided by an accelerometer that had been taped to the hoof and the signal from the accelerometer was recorded simultaneously with the output of the flash contact of the camera, thus permitting synchronization. Truncated Fourier series represented the kinematic data obtained.

As could be expected, in the proximal parts of the limbs much larger displacements were recorded than in the distal parts. Displacements along the X-axis varied from 18mm (proximal part of the scapular spine to 71mm (caudal part of the greater trochanter of the femur). Displacements along the Y-axis (perpendicular to the long axis of the bone) varied from 6mm (proximal radius) to even 146mm (caudal part of greater trochanter) (van Weeren et al. 1990a,b).

Two-dimensional analysis of skin displacement – correction models

Based on the experimental data, correction models were developed for the skin displacement artefact (van Weeren et al. 1992a). These were based on
obtaining the estimated 2D displacement vectors \( r_i = (x_i, h_i)^T \) for each landmark \( i \), expressed in a coordinate system fixed to the corresponding bone, for each sample of a recorded movement. Then, a coordinate system was defined using two easily palpable landmarks with the direction of the bone axis parallel to the line connecting these landmarks. Displacements were normalized to fractions of the distance between the two bone landmarks (fig. 5). When the coordinates \( r'_i = (x'_i, y'_i)^T \) of two skin markers \( i = 1,2 \) on the same bone are known, and predicted normalized skin displacements \( r_i \) are available, the coordinates \( r_i = (x_i, y_i)^T \) of the corresponding bone landmarks can be solved from:

\[
  r'_i = r_i + R(a). r_i. |r_1-r_2| \quad (i = 1,2)
\]

where \( R(a) \) is the rotation matrix corresponding to the orientation of the bone (van den Bogert et al. 1990).

Two-dimensional analysis of skin displacement – example of application

The correction models were used to try to explain an observation that thus far had not been understood. In the horse, the movements of the stifle and hock joints are coupled through two ligamentous structures that bridge both joints. These are the peroneus tertius tendon at the cranial side and the almost entirely ligamentous superficial digital flexor muscle at the caudal side. This so-called reciprocal apparatus makes that a flexion of the stifle joint is always accompanied by a flexion of the hock joint and vice versa. When trying to actually measure this coupling using kinematic data it appeared, however, that the coupling was much less strict than could be expected on anatomical grounds, especially in the second half of the stance phase (Wentink 1978). The explanation given was that perhaps kinetic energy was stored as elastic energy during this phase of the stride, which would help in flexing the limb during the early part of the swing phase. A similar mechanism is known to occur in the superficial digital flexor tendons of the horse (and the Achilles tendon in man). However, when the measured angles were translated into tendon strain, more than 20% strain would be present, which is far beyond the physiological limits. Therefore, such a mechanism could not be the only explanation.

The application of the newly developed correction models for skin displacement solved the mystery. In fact, the skin displacement artifact caused the apparent flexion of the stifle during late stance. When corrected for this artifact, the joint angle appeared to remain stable throughout the entire stance phase, which is in line with the angular pattern of the hock angle (Fig. 6a). Interestingly, when plotting Krüger’s data from the late 1930s (Krüger 1938), which were obtained using the oiled skin technique, a similar curve (Fig. 6b) was obtained as when using the skin displacement correction models (van Weeren et al. 1990c). When using the corrected angles, there still appeared to be storage of elastic energy in the peroneus tertius tendon, but much less than when using uncorrected angles. This was later confirmed by direct measurements in the tendon, using implanted mercury-in-silastic strain gauges (van Weeren et al. 1992b).
Conclusion and further developments
The skin displacement artefact is an important confounding factor in kinematic studies using skin-fixated markers. If these studies only have a comparative nature, correction may be omitted because the artefact can be supposed to affect horses of a similar conformation to a similar extent. When, however, the exact nature of the movement of skeletal structures is the target of study, some way of either avoiding or correcting for this error is imperative.

Different approaches for skin movement artefact reduction have since been developed. In most situations, the skeleton can be modelled as a series of linked segments, and this additional information causes the joint angles to be less sensitive to skin marker artefacts (van den Bogert et al. 1994). These methods were a precursor to modern model-based global optimization methods for kinematic analysis (Lu and O’Connor 1999, Roux et al. 2002). Some work on (3D) skin displacement in the horse has been done in recent years (Khumsap et al. 2004, Sha et al. 2004), but these latter more advanced methods have not yet been applied in equine movement studies.

References


Legends for illustrations

**Fig. 1** The set-up at Palo Alto (California) where Muybridge did his first investigations of animal locomotion using a linear array of cameras (1878).

**Fig. 2** Serial photographs of a trotting horse. The suspension phase has been visualized for the first time in history.

**Fig. 3a** Schematic drawing of a LED glued in a stainless steel tube for implantation in the bone underlying the area of the skin under investigation.

**Fig. 3b** Schematic drawing of cross-section through the equine 3rd metacarpal bone showing the position of the drill and of the implanted LED.

**Fig. 3c** Close-up of fetlock area showing one large and 4 small skin markers with the LED in the center of the latter shining through the skin.

**Fig. 4a** Schematic drawing of the equine femur with Steinmann pin and plastic strip equipped with 2 reference markers. a: strip with reference markers; b: bone axis; c: trajectory of skin marker on the cranial part of the greater trochanter; d: trajectory of skin marker on the lateral epicondyle; x,y: axes of the orthogonal coordinate system.

**Fig. 4b** Photograph of horse equipped with Steinmann pin and plastic strip.

**Fig. 5** Schematic drawing of the hind limb of the horse, showing the femur coordinate system , positions r'₁ and r'₂ of the skin markers (o) and the positions r₁ and r₂ of the corresponding skeletal landmarks ( ).

**Fig. 6a** Curves of stifle angle as obtained by cinematographic recording without (continuous line) and with (dashed line) correction for skin displacement.

**Fig. 6b** Curves of stifle angle without correction for skin displacement as obtained by cinematographic recording without (continuous line) correction for skin displacement and as obtained by the oiled skin technique employed by Krüger (1938) (dots). Note the resemblance of the last curve with the corrected curve in fig. 6a.
Fig. 4b
Fig. 6b
SESSION 6:

Mocap and Animation
Motion capture for new media - then and now

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The attributes of motion capture that are well understood by most practitioners in new media applications are those of realism and productivity. Motion capture data provides the most realistic data that can be attained while also providing extremely high levels of productivity. Thus it's not surprising that video game manufacturers interested in producing sports games were among the first to adopt motion capture and use it widely. The visual sense and the business sense of the process were both well met.

More artistic uses of motion capture, however, have been slower to adopt motion capture as a primary tool. Some of the reasons for this will be discussed, but the principal one that will be addressed is the notion of feedback. That is, the way a tool provides information back to the user. For an artist to explore an expression space they need quick turnaround in their results. This has frustrated many traditional film directors who are used to getting an immediate sense of what is happening in a staged environment.

A motion capture animated short that was produced 9 years ago, using tools available at that time, will be compared to a "making of" video produced last month describing the future of film production using motion capture. The differences in technology will be highlighted and used to describe the arc of innovation over the last decade. A special emphasis, however, will concentrate on what is the same about these two productions and talk about why the time is now finally ripe for motion capture to make a dramatic impact on the world of film making.
Motion Capture and Animation

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Animal motion capture and animation is currently used for three main purposes: scientific research, veterinary investigation and in the film and game industry. The animation of motion plays a pedagogical and demonstrative role in scientific and medical applications of motion capture but the entertainment industry is unique in having animation rather than analysis of motion as a primary aim. The following discussion therefore focuses on film industry capture and animation.

Capture

Capture within a research environment has the secondary constraint of cost effectiveness. Minimum numbers of cameras and the use of video cameras rather than dedicated digital optical systems introduce a new layer of technical problems to be tackled. Film industry applications encounter new problems such as the need to capture extremely large volumes, trials with up to ten subjects and brightly lit location shoots, but operate on a larger budget allowing larger set ups to be explored with up to 36 near infra-red cameras recording a single take.

Currently optical systems still dominate motion capture, with magnetic systems yet to perform and inertial systems just beginning to attract attention. Optical systems have the advantage of being wireless but currently still require the attachment of markers to landmarks on the subject. Whilst research applications prefer to attach markers directly to the subject in the film industry lycra suits are more common, when the climate allows, with the markers being attached by velcro. These are generally well tolerated by animals and greatly increase the staying power of the markers. The large amount of cameras allows the use of high numbers of small, soft markers which help to reduce problems due to suit movement and inertia of the markers. Motion capture suits are often tailor-made for the subject, and a unique marker set allows automated differentiation between actors in multiple actor capture. Whilst black is the preferred suit colour blue screen suits are necessary when combining motion capture with simultaneous traditional filming, to allow the actor to be removed from shot easily.
Modeling & skeletons

The skeletons used to convey the motion are link segment models with a pelvis based root. The skeleton is globally optimised to remove skin movement and customised to the individual. Skeletons need to be robust enough to create an approximate animation in real time during filming, aiding actors who are playing creatures with unusual morphology, and anyone interacting with them, to create a more realistic performance. The skeletons used are not anatomically accurate other than in their joint constraints, which constitute the majority of the skeleton design. Each joint needs to be rigidly constrained in order to map the motion of the subject capture onto the final size and shape of creature required whilst maintaining a realistic motion. Unlike clinical gait analysis in the film industry any abnormalities need to be removed. Extra routines ensure that the feet stay on the floor during stance and simulate the movement of any missing markers, extra limbs, heads or tails.

Figure 2 Vicon IQ screen shot
Figure 3 Example “skeleton” and skin

Skinning

“Skinning” for animation represents the addition of muscles and hair to the skeletal model. A three dimensional wire mesh will be dynamically tethered to the skeleton, allowing the movement of the skeleton to be translated into skin and muscle movement. The potential use for this technique as a tool for removing skin movement has yet to be tested. Eventually the addition of textures such as hair, or a scan of the subject’s face will complete the final character, which will then need rendering, lighting, shading and compositing into the final scene.
Figure 4 Example final animation.
Software
In the film industry as elsewhere, proprietary software and in-house programmers are becoming the norm; however commercial packages are still present including Vicon IQ, which differs subtly from Workstation, 3d Studiomax, Motionbuilder and Maya.
Applications of Motion Capture Technology in Veterinary Medical Education

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Motion capture, animation and interactive three-dimensional simulations have the potential to revolutionize the way in which students learn to recognize normal and abnormal gaits. Until quite recently, students in the classroom environment were dependent on two-dimensional illustrations and textual descriptions in order to understand gait patterns. The limitations of this medium were quickly realized. The advent of affordable digital video systems, common in today’s clinics and classrooms, now allow veterinarians to record actual cases and convey subtleties much more accurately. Video is somewhat more effective, but it still has limitations. Data that were not recorded by the video camera, for example from the other side of the animal, are not available to the student. The person watching the video clip is unable to interact with the data, other than using basic playback controls. More specifically, they cannot change their viewing perspective, which is something they’d do naturally if they were observing the animal in a real-life situation. Video, while an improvement over illustrations, can still feel artificial and is far from being truly immersive.

Motion capture, if used in combination with other educational technology products, can overcome these limitations. One of the major advantages of using motion capture rather than video is that data can be recorded from the entire animal. Since the data exists in a three-dimensional environment, viewing perspectives can easily be changed by creating virtual cameras in the system. The learner is no longer limited to the camera’s point of view; the subject can be observed from the side, behind, or even below. Motion data can also be “cleaned” to optimize the experience. Clips can be slowed down, sped up, or placed in combination, one after another. The possibilities that exist in a virtual environment are almost endless.

Several off-the-shelf software tools are available to streamline data processing. Products such as Alias MotionBuilder, 3DS MAX and Maya are capable of manipulating data from motion capture systems and applying it to 3D models. These packages have become very affordable and run on typical computers and operating systems.

Programming tools, such as Macromedia Director, allow designers to create truly interactive three-dimensional simulations. The objective is to use motion capture data in context, giving the user complete control over the virtual environment. Rather than passively watching an animation created from motion capture data, which still has some of the limitations of video, the user can now have complete control over the camera. While the animal is moving, the virtual camera can be dynamically repositioned to obtain a new perspective. This more closely mimics a real-life
interaction in some respects (moving toward/away, from side to back) but also adds “superhuman” capabilities such as hovering above the subject or tracking a specific limb.

Motion capture and animation will have an enormous impact on veterinary medical education. Much like the digital revolution has changed diagnostic imaging, the process of teaching gait patterns will never be quite the same. With the combined power of real-life data and creative programming, the experience could someday rival real-life patient encounters.
SESSION 7:

Medical Applications of 3D Analysis
Medical Applications of 3-D Movement Analysis

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Introduction

Three dimensional movement analysis can produce clinically relevant information on joint kinematics (movement) and joint kinetics (forces and moments). Here we will present an overview of the potential clinical uses of such information, with specific examples from orthopaedics, neurology, and sports medicine.

Kinematic variables

Joint rotations are useful descriptors of functional deficits and locomotor performance. Movement patterns can be used for diagnostic purposes (comparison against normal controls) or for monitoring of therapies (comparisons to the patient’s own data). These interpretations are mainly statistical rather than mechanical, but statistical methods are more powerful when a few kinematic variables are carefully selected based on a mechanistic understanding of the clinical problem. For instance, in the study of anterior knee pain, studies have focused the internal-external rotation of femur and tibia, as these variables can affect patellofemoral tracking [6].

Joint rotations can be input into a musculoskeletal model, such as SIMM (Musculographics, Inc., Chicago, IL) in order to compute length changes of muscles during movement. In cerebral palsy, such information can identify muscles with excessive stiffness that should be considered for surgical interventions [1]. In horses, where many muscles have short (or no) fibers, this type of analysis can provide remarkably accurate information on tendon strain and force [7][8]. Ligament function in humans can potentially be evaluated with the same techniques, but the length changes in short ligaments are too small to be reliably estimated with kinematic methods.

Kinetic variables

Joint moments are typically interpreted as being generated by muscles, but this is typically only true for those axes in the joint coordinate system (JCS) where the joint has a large range of motion. For instance, the “abductor” moment in the human hip joint clearly represents muscle function. But the same variable in the elbow during throwing should be interpreted as tension in the ulnar collateral ligament. These non-muscular kinetic variables are mostly used in orthopaedics, as they indicate loads on connective tissue: cartilage, ligaments, and bone. For instance, external valgus moment in the knee is a prospective risk factor for ligament injury in soccer and basketball [4]. Knee varus loads are associated with medial compartment osteoarthritis [5]. The muscular joint moments are typically more relevant for neurological evaluations or for sports performance. If quantitative estimates of muscle forces are required, muscular joint moments must be separated into the
contributions from individual muscles. Optimization methods have been well developed [2] but have not found widespread use in clinical movement analysis.

Joint forces obtained from a standard kinetic analysis reflect the resultant of all forces crossing the joint. As there are both compressive forces (articular contact) and tensile forces (ligaments and muscles), this resultant is always much smaller than the forces in the actual anatomical structures. Estimation of articular contact forces will therefore always require that individual muscle forces are also estimated. The same is true for estimating ligament forces, if there are muscles that have a significant agonistic or antagonistic function relative to the ligament. For example, the external resultant force at the knee is usually directed posteriorly to the tibia. But if the anterior pull of quadriceps is taken into account, the load placed on the passive structures is anterior, indicating tension in the anterior cruciate ligament, rather than the posterior cruciate ligament.

Joint power is obtained by multiplying a muscular moment by joint angular velocity. In three dimensions, the dot product of the two vectors provides the total power of all degrees of freedom. If separate estimates for the different JCS axes are required, this would require that joint moment is quantified on each JCS axis, rather than in bone coordinates as is the common practice. Joint power is often an overestimation of actual muscle power, because the analysis does not take into account that some major muscle groups act across two joints. This is even more of a problem in horses, and it is doubtful whether useful clinical interpretations of joint power are possible.

**Clinical research and patient care**

Movement analysis can be applied in clinical research as well as in patient care. Research applications include retrospective designs, prospective designs, and clinical trials. Successful applications in human sports medicine were related to patellofemoral pain, stress fractures, and anterior cruciate ligament injuries.

In patient care, potential applications are in diagnostics and in follow up evaluations. Cost reimbursement for clinical movement analysis is still problematic, and this has discouraged the development of clinical applications. Modern technology and software has the potential to significantly decrease the cost of clinical movement analysis, but we still require solid cost-benefit analyses before movement analysis can become a routine tool in patient care. A promising technological advance is the advent of real-time tracking of passive markers. This will not only reduce the time needed to generate the report of the analysis, but also has potential as a biofeedback tool for gait retraining in physical therapy [3]. Here, the technology is not only used for evaluation of treatment, but has become an integral part of the treatment itself.

**References**


SESSION 8:

Applications of 3D Kinematics
A new method for analyzing symmetry and normalcy of gait patterns is explained. This method utilizes eigenvectors to compare waveforms. The right and left limbs of a single subject are compared to determine symmetry. To determine normalcy, a single limb from a subject is compared to a normative file created from the average of healthy control subjects. The analysis method provides four measures of symmetry and normalcy: trend phase, trend symmetry/normalcy, range amplitude ratio, and range offset. The trend symmetry/normalcy is a comparison of the shapes of the waveforms for each limb. The range amplitude ratio is a comparison of the range of motion of each limb. The range offset is a comparison of the range in which each limb operates. The exact methods and clinical applications are provided.
Articulated Figure Skating Boots Can Reduce Impact Forces

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Figure skating has evolved into a highly specialized and athletic sport with difficult triple and quadruple jumps increasingly becoming the primary focus. Skaters are spending more time practicing jumps and overuse injury rates show an associated increase. The current rigid leather skating boot severely restricts ankle motion, reducing the body’s ability to absorb damaging landing stresses. Modifying the current skating boot by introducing an ankle articulation may help reduce peak transient forces during landing. The purpose of this study was to test the ability of an articulated skating boot to attenuate ground reaction forces during landing. It was hypothesized that articulated figure skates would reduce peak forces as well as loading rates when compared to traditional boots. Prototype articulated boots were manufactured by Jackson Ultima Skates inc. Nine US Figure Skating juvenile-level or higher figure skaters from the surrounding area were tested in off-ice simulated jump landings from a 30cm platform onto a force plate. Kinetic and kinematic data were collected and analyzed from skaters wearing standard skating boots and, after a brief training period, wearing articulated boots. A repeated measures ANOVA was used to determine significant differences (alpha=.05) in ground reaction forces, loading rates, and several kinematic variables. Peak heel impact forces and loading rates decreased significantly in the articulated boot. Ankle plantarflexion and boot/ice angles at impact increased while the total joint flexion at impact decreased. There were no statistical changes in toe impact force, jump height, and hip and knee angles at impact. The majority of skaters used the increased range of motion in the articulated skates to effectively reduce landing forces and loading rates. However, a few showed very little change between conditions, even with increased ankle plantarflexion and boot/ice angles at toe contact. In some skaters, the ankle musculature was apparently not utilized to stiffen the ankle during the impact phase. The findings suggest that some skaters attempt to utilize the ankle, while others do not, and that some retraining may be necessary to decrease landing forces.
SESSION 9:

Inertial Sensors for Motion Capture
Motion Capture Using Inertial Sensing

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Introduction

Determining position, velocity and rotation rates of an object using accelerometers and gyroscopes is an established field in navigation and guidance of aircraft, marine vessels and missiles. Primary advantage of these systems is total independence on external systems and a high-frequency response. Because the principle of both gyroscopes and accelerometers is based on the resistance of a moving mass to a change in motion they are also referred to as inertial sensors. Traditional navigation systems are accurate, but expensive, large, bulky and can not be body worn. Since the '90s micromachined accelerometers and gyroscopes have become available for the automotive industry (airbags) and consumer electronics (steady shot cameras). These developments enabled 3D orientation tracking using relatively low-cost matchbox sized sensors (Foxlin et al, Bachmann, Luinge et al). Inertial Measurement Units (IMU) output orientation without line of sight restrictions and without the need for a fixed lab setup. Markets for IMUs include sports, rehabilitation and entertainment to name a few. This abstract will shortly explain the principles of an IMU and how it can be used in human/equine motion capturing.

Signals with respect to sensor casing

An IMU contains the following components:

1. Three accelerometers measure acceleration and gravity along different sensitive axes. Typical ranges are 5g and 10g.
2. Three gyroscopes measure angular velocity using the coriolis effect. Typical ranges of small size micro machined gyroscopes go up to 1700 deg/s.
3. The (earth) magnetic field is measured using three magneto resistive magnetometers.

The 9 sensor components are mounted orthogonally in a box and calibrated for temperature dependencies, direction of sensitive axes w.r.t. sensor casing, gains and offsets.

Orientation

Orientation can be obtained by mathematically integrating the angular velocity starting at a known initial orientation. Due to small errors in the measured angular velocity, the obtained orientation will contain an ever-increasing error. This is prevented by considering the direction of gravity measured by the accelerometer. A Kalman filter has been developed that fuses the orientation measured by gyroscopes with the orientation measured by accelerometers and magnetometers yielding a drift free, high fidelity orientation measurement. Typical errors are smaller than 2 degrees rms under dynamic conditions.
Position

In much the same way as with orientation the acceleration can be used to obtain velocity and position. Because of the accumulating integration drift due to small measurement errors, additional measurements or assumptions are required to limit this integration drift to acceptable limits. For human motion capture an articulated body can be used, obtaining position by adding segments of known length and orientation. Other possibilities include the use of GPS, externally/sensor mounted cameras or knowledge about the movement itself. The latter is demonstrated in figure 1, where the velocity is obtained by assuming that the hoof was still on the ground in between strides.

Figure 1. Kinematics of left frontal hoof while slowly trotting on a rubber mat. Top: Angular velocity in lateral (-) and vertical (..) direction. Middle and bottom: Acceleration resp. velocity in trotting direction (-) and vertical direction (..).

Conclusion

Inertial sensing to accurately measure orientation is a proven method. Although some issues remain to be solved, the application of inertial sensors in full body motion capturing is promising because of its ambulatory nature and direct measurement of angular velocity and acceleration.

References

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