Quantification of equine stifle passive kinematics

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OBJECTIVE
This study aims to quantitatively characterize the passive kinematics of the healthy, soft tissue-intact equine stifle to establish an objective foundation for providing insights into the etiology of stifle disorders and developing a possible surgical treatment for stifle degenerative disease.

ANIMALS
5 whole-horse specimens.

PROCEDURES
Reflective markers with intracortical bone pins and a motion capture system were used to investigate the stifle joint kinematics. Kinematics of 5 whole-horse specimens euthanized within 2 hours were calculated for internal/external rotation, adduction/abduction, and cranial/caudal translation of the medial and lateral femoral condyles and estimated joint contact centroids as functions of joint extension angle.

RESULTS
From 41.7° to 121.6° (mean ± SD, range of motion: 107.5° ± 7.2°) of joint extension, 13° ± 3.7° of tibial external rotation and 6° ± 2.7° of adduction were observed. The lateral femoral condyle demonstrated significantly greater cranial translation than the medial during extension (23.7 mm ± 9.3 mm vs. 14.3 mm ± 7.0 mm, \(P = .01\)). No significant difference was found between the cranial/caudal translation of estimated joint contact centroids in the medial and lateral compartment (13.3 mm ± 7.7 mm vs. 16.4 mm ± 5.8 mm, \(P = .16\)).

CLINICAL RELEVANCE
The findings share similarities with kinematics for human knees and sheep and dog stifles, suggesting it may be possible to translate what has been learned in human arthroplasty to treatment for equine stifles.

Osteoarthritis (OA) is a common cause of significant lameness in equine stifles.\(^1,2\) It was reported that OA in equine stifles predominantly occurs in the medial compartment of the femorotibial joint and cartilage damage on the medial femoral condyle is often recognized as an early sign of stifle OA.\(^1,3\) Thus, a significant amount of research into cartilage healing has been completed for equine OA treatment such as mesenchymal stem cell (MSC) administration and autologous platelet concentrates (APC).\(^4,5\) It was reported that 27% of the horses with fibration or small areas of damage in the femorotibial joint failed to return to work after the treatment with arthroscopic MSC.\(^4\) Moreover, horses with OA in various joints treated with APC injection showed improvement in lameness scores and joint effusion, but there was insufficient evidence to conclude that APC can be used for equine OA treatment.\(^5\) Therefore, there is a strong clinical motivation to develop novel treatments for equine stifle OA as a pain-relieving, functional, and life-sparing treatment.

A common and successful treatment in humans for isolated medial compartment OA with intact cruciate ligaments is medial unicompartmental knee arthroplasty (mUKA).\(^6\) It is intuitively appealing to think an equine mUKA could be a feasible treatment and provide the same benefits: pain relief, restoration of joint function, range of motion, and athletic ability. To develop a unicompartmental stifle arthroplasty technique and design the implants, a fundamental understanding of the kinematics of the equine stifle joint is needed.\(^7,8\) Skin marker-based motion capture is a common method to investigate human knee kinematics as well as to analyze equine stifle kinematics. It was reported that the equine stifle displayed approximately 35.1° of flexion/extension during walking ranging from 126.9° to 162° at maximum extension (in veterinary terms, full joint extension is described as 180°).\(^9\) However, skin displacement between the markers and the underlying bone structures, especially at the proximal region of the equine hindlimb, could result in errors up to 15° in
joint angles. One study utilized radiostereometric analysis (RSA) to investigate the 3-dimensional (3-D) equine femorotibial joint kinematics. However, important soft tissues and joint structures such as muscles, joint capsule, and patella, which are known to contribute to joint functional stability, were completely removed. Therefore, there is currently insufficient information on the equine stifle kinematics to define the performance criteria required to guide the design of an equine mUKA.

Motion capture with reflective markers fixed to intracortical bone pins was considered as the “gold standard” for investigating human knee kinematics in vivo as the pins were inserted directly into the bone structures, providing joint kinematic data that were free of skin displacement errors. Therefore, this study aims to describe the passive kinematics of the healthy, soft tissue-intact equine stifles using bone pins and a motion capture system to establish an objective foundation for developing surgical therapies for equine stifle degenerative disease. We expect the healthy equine stifle will demonstrate a similar pattern of tibial external rotation and abduction followed by adduction throughout the extension as reported in other species such as humans, sheep, and dogs. We also expect the lateral compartment of the femorotibial joint will display a greater craniocaudal translation in magnitude than the medial femorotibial compartment.

Materials and methods

Specimen preparation

Horses euthanized for reasons unrelated to joint disease were used for this study with appropriate institutional approvals. To minimize the possible alterations to the stifle kinematics induced by rigor mortis, all tests were performed within 2 hours after euthanasia. One side of the hindlimbs was randomly chosen using a coin toss to be evaluated. Animals were positioned in dorsal recumbency on a surgical table and the hindlimbs were attached with nylon straps at the pastern to an overhead cable and motor system to passively extend and flex the limb without restricting motion. The lateral femoral epicondyle, proximal tibial tuberosity, and the cranial proximal base of the patella were chosen to be the pin insertion sites as they would not affect or tether soft-tissue mobility throughout the joint range of motion (Figure 1). To minimize the impingement between the skin and bone pins, approximately 3-mm stab incisions down to the bone were made using a number 10 blade along the estimated direction of translation of the skin at each intracortical pin insertion site.

Motion capture equipment and augmented tools

To guide the bone pin insertion, 3 rigid rectangle blocks (20 mm X 40 mm X 26 mm) each with 2 converging pin holes were designed in SolidWorks (Dassault Systèmes SolidWorks Corp) and 3-D printed (Fortus 450mc; Stratasys) with acrylonitrile butadiene styrene (ABS). The 3-D-printed blocks also served as mounts for marker posts and retroreflective markers. In each block, two 4-mm-diameter threaded Steinmann pins (Zimmer Biomet) were inserted into the bone convergently through the pin holes. Three rigid marker arrays each with a unique quadrilateral-shaped top and a hollow rectangle base (same dimension as the block) were also designed and 3-D printed with ABS. On each array, 4 M4 threads were created at each corner of the quadrilateral top to provide fixation for ABS. Two different lengths of the posts, 25 mm and 50 mm, were used for optimal marker spatial distribution as marker overlapping could result in position

Figure 1—3-D demonstration of intracortical bone pin placement on a right stifle. A—Caudal view of 2 pins inserted at the lateral femoral epicondyle. B—Lateral view of the pin insertion sites at the proximal tibial tuberosity. C—Lateral view of the pin insertion sites located at the cranial proximal base of the patella.
ambiguity throughout the range of motion. Four 15.9-mm-diameter retroreflective markers with M4 threads were attached to the marker posts on each array. The marker arrays were mounted on top of the corresponding blocks and fixed with 2-mm-diameter Kirschner wires (Figure 2).

**Figure 2**—Demonstration of femur, tibia, and patella marker arrays each with 4 reflective markers. Each array was guided and fixed by 2 converging intracortical bone pins.

**Kinematic data collection and coordinate system establishment**

Markers were observed with 4 high-speed cameras (Motion Analysis Corporation) using a calibrated motion capture system collecting data at 200 Hz. A minimum of 3 cycles from the maximum angle of passive flexion to extension at a uniform speed of 0.2 m/s were collected with 1 cycle taking approximately 18 seconds. The motor was controlled by the same researcher using visual cues to estimate the maximum flexion angle where the limbs stopped further flexion without any external forces (Figure 3). The maximum extension angle was reached before the patellar locking occurred. After kinematic data collection, marker arrays were left in situ and the tested limb was disarticulated at the coxofemoral joint. CT evaluation was performed using a 160-slice Toshiba Aquilion CT Scanner (Cannon Medical System) and helical volume data (slice thickness of 0.5 mm and 0.3 mm slice overlap) was acquired. The bone algorithm was used for all CT-based 3-D reconstructions and analyses. Single energy metal artifact reduction (SEMAR) was implemented on the slices that included the bone pins. The segmentation of 12 spherical markers and the bones including femur, tibia, and patella was performed in Mimics (Materialise) based on the CT images using −500 to 3,132 Hounsfield units (HU) and 226 to 3,132 HU, respectively. The damaged bone surface resulting from bone pin insertions was reconstructed for optimal visualization. Segmented 3-D models were exported to 3-Matic (Materialise) where geometric relations between the anatomical landmark-based coordinate systems and the markers were determined. To preserve the native anatomic geometry of the joint as much as possible, the use of the smoothing function was minimized and avoided at the regions of anatomical landmarks that were used for creating coordinate systems. Spherical approximation to each marker was created, and with 3 randomly chosen centers of the markers on each array, 3 Cartesian coordinate systems were determined. To establish coordinate systems for each bone, a method similar to what is reported in human knees was used. To minimize variation in choosing anatomical landmarks, all coordinate systems were created by one researcher for all specimens (Figure 4).

**Kinematic data processing**

Kinematic data were filtered with a zero-phase fourth-order Butterworth low-pass filter having a
The cutoff frequency of 0.3 Hz (MATLAB; The MathWorks Inc). Femorotibial kinematics were calculated using standard conventions for translations and rotations.\textsuperscript{21} The kinematics were resampled to 101 points/cycle and then averaged (mean) across 3 cycles for each specimen. Cranial-caudal (CrCa) motion of the femorotibial joint was described as the translation of the centers of the spherical approximation to the femoral condyles. Estimated contact centroids on the medial and lateral tibial plateau were also calculated using a program that evaluated at the distances of all the nearest neighbor points on both surfaces of the femoral condyles and the tibial plateau, finding and then averaging the coordinates that correspond to the
distance that are less than the assigned contact cut-off value. As a previous study\textsuperscript{22–24} reported the average articular cartilage in 5 locations on the femoral trochlea and medial femoral condyles in the equine stifle was estimated to be 2.2 mm, with the thickness of the meniscus taken into account, the contact cut-off for the program was set at 8 mm. Individual data curves and group means were plotted for femorotibial internal/external (IE) rotation, abduction/adduction and CrCa translations of the medial and

Figure 5—The passive kinematics of the equine stifle showing tibial external rotation (A), adduction (B), caudal medial (C), and lateral (D) condylar translation (centers of the spherical approximation) and estimated contact centroids translation on the medial (E) and lateral (F) tibial plateau as functions of stifle extension angle. Each graph shows the group mean (thick dark blue line) with ± 1 SD (light blue shaded area) in addition to the individual joint data with each thin line in a different color representing the dataset calculated for each specimen.
lateral femoral condyles as well as the estimated joint contact centroids on the tibial plateau as a function of femorotibial joint extension angle.

Statistical analyses

Means and standard deviations were calculated for femorotibial joint kinematics including flexion/extension, IE rotations, adduction/abduction, and translation in CrCa direction. Femoral condylar translations measured with the spherical approximations as well as the estimated joint contact centroids in the medial and lateral compartments were compared using Student’s paired t tests. Differences were considered statistically significant when \( P < .05 \).

Results

To minimize human errors and despite attempts to avoid damage to arrays, specimens with markers damaged or disturbed during the limb harvesting or transporting after the kinematic testing were excluded. Nine animals were used in total, although 4 were excluded due to array damage. Thus, only 5 datasets were reported in the current study. The 5 specimens reported were all Thoroughbreds, 3 geldings and 2 mares, with a body mass of 501.2 ± 40.1 kg and 9 ± 3.5 years of age.

Femorotibial joint rotations

All flexion/extension angles reported were in veterinary terms. The primary rotational motion of the femorotibial joint was flexion/extension, ranging from 41.7° flexion to 121.6° extension with a mean of 107.5° ± 7.2°. As the joint extended from flexion to maximum extension, the average peak-to-peak displacement of tibial IE rotation for 5 specimens through range of motion was 13.0° ± 3.7°. The group mean ranged from −14.4° in internal rotation to −2.3° as the joint extended (Figure 5). All tibiae demonstrated continuous adductions during joint extension averaging 6.0° ± 2.7°, with the group mean ranging from −6.3° in abduction to −0.4°.

Femorotibial joint translations

The center of the best-fit sphere to the lateral femoral condyle demonstrated a significantly \( (P = .01) \) greater CrCa translation than the medial condyle throughout the range of motion, averaging 23.7 mm ± 9.3 mm and 14.3 ± 7.0 mm, respectively. The group mean displayed a 11.1 mm peak-to-peak medial femoral condylar translation from −18.4 mm (negative value meaning femoral origin being caudal to the tibial origin in tibial coordinate system) to −7.3 mm, and the lateral femoral condyle translated from −33.6 mm to −9.9 mm, thus 23.7 mm cranial (Figure 5). However, the difference between the CrCa translations of the estimated joint contact centroids on the medial (13.3 mm ± 7.7 mm) and lateral (16.4 mm ± 5.8 mm) tibial plateau was not statistically significant \( (P = .16) \). The group mean showed a 11.3 mm (∼5.0 mm to 6.3 mm) and 15.9 mm (∼12.9 mm to 3.0 mm) cranial translation during extension in the medial and lateral compartment, respectively.

Discussion

To our knowledge, this is the first report of 3-D femorotibial passive kinematics in the fully intact equine stifle. The bone pins mounted with markers provided direct and sensitive kinematic data of the joint without any soft tissue removal. The current study demonstrated IE rotation, adduction/abduction, and CrCa translation of the tibia were tightly related to joint flexion/extension angle, which is similar to what has been reported in human knees, sheep and dog stifles.\(^ {15,17-19} \) The equine tibiae experienced approximately 13° of external rotation and 6° of adduction throughout the joint extension. The best-fit approximations to the femoral condyles demonstrated the lateral femoral condyle experienced greater translation than the medial femoral condyle by 12.6 mm, which agrees with the results of the estimated joint contact centroid translations on the tibial plateau as the lateral contact centroid translated 4.6 mm greater than that of medial. However, no statistical significance was found in the difference in estimated contact centroid translation. One possible explanation for this is anatomical. The contact centroid estimation does consider the curvature of the tibial plateau. The lateral tibial plateau has a relatively steeper slope caudally in comparison to the medial compartment, resulting in the estimated contact centroids being more cranial than the center of the femoral condyle. The mean axial center of rotation, or pivot location, was extra-articular to the medial side of the joint given both femoral condyles moved cranially with extension, but more in the lateral compartment than in the medial (Figure 6).

There is a paucity of literature on equine femorotibial joint kinematics, the only comparable study was by Halley et al.\(^ {12} \) Their study performed an RSA on the equine femorotibial joint kinematics and reported rotational motions of the joint as well as the estimated articular contact patterns within the range of 110° to full extension. The present report corroborates others\(^ {12} \) which demonstrate external rotation and cranial translation in both medial and lateral compartments during joint extension. Significantly higher CrCa translation on the lateral tibial condyle than on the medial was also observed in Halley et al.\(^ {12} \) However, some differences were seen in methods and results. First, the specimens used in Halley et al were dissected free of muscles, which are vital to the femorotibial joint kinematics. One study\(^ {25} \) described the equine popliteus muscle and tendon, which originates from the lateral condyle of the femur and inserts on the caudomedial border of the tibia has a constant attachment to the lateral meniscus, suggesting the popliteus could contribute to the stifle IE rotation and translations in addition to stabilizing the lateral meniscus. Although further investigations are needed to determine the exact
effect of equine muscular structures on the stifle passive kinematics, it is appropriate to assume the preservation of the muscular structures and the joint capsule would be optimal. Second, the current study reported a higher maximum flexion angle and lower maximum extension angle than that of Halley et al. This was due to the differences in specimens used in the studies. In Halley et al., the femorotibial joints were dissected free of patella and its associated soft tissues, including medial, middle, and lateral patellar ligaments and quadriceps, thus patellar locking was not a concern in high joint extension angles. While the current study utilized fully intact hindlimbs, and due to lack of knowledge on detailed patellar locking angle, maximum femorotibial extension was visually estimated to be before patellar locking, thus a lower maximum extension angle. Third, Halley et al. observed an approximately 3.5° of tibial abduction from 110° to 135°, followed by 3° of adduction from 135° to full extension. In contrast, in current study, there was no tibial abduction displayed throughout the joint extension. This may have resulted from the external force exerted by the strap and motor, where the limb was pulled vertically and medially by the strap aligned with the centerline of the body. Finally, the current study reported a higher CrCa joint translation throughout the range of motion than that of Halley et al. One possible explanation was lack of compressive force from the femoropatellar joint. Although the impact of the compressive force in the femoropatellar joint on the femorotibial translation during flexion in equine stifle was not evaluated, it was reported that the thickness and the positioning of the patella in human knees can affect femorotibial IE rotation and anteroposterior translation, especially during flexion. Higher posterior translation in the lateral femoral condyle was observed in the case of thicker patella compared to thinner patella, suggesting lack of patella could result in reduced lateral femoral condylar posterior translation during flexion.

One limitation of the current study was the absence of active muscle forces and realistic joint loading. Due to the dorsal recumbency position, the hindlimbs were manipulated through a vertical

Figure 6—Trajectory of translation of the estimated femorotibial joint contact centroids on the tibial plateau in axial view during joint extension at 35° (A), 60° (B), 110° (C), and 125° (D). The red circles in each image are the estimated joint contact centroids at each time frame. The orange star in D represents the mean center of axial rotation.
force at the distal tibia by a motor system; thus, our loading pattern could potentially alter the joint kinematics. The flexion/extension and IE rotation at the coxofemoral joint through the motion cycles could have played a role in limiting the stifle range of motion. Specifically, during the coxofemoral and stifle extension, semitendinosus and semimembranosus were elongated at the tibial insertion site, which limited further stifle extension. However, utilizing a standardized experimental setup with specimens in dorsal recumbency and manipulated using the motor system enhanced inter-specimen repeatability. The disruption of the soft tissue function due to bone pin insertion or the stab incision is another limitation of the current study. Although the pins in the tibial tuberosity and the patella were created through stab incisions in the skin, they could still place tension on the skin as it slides over the underlying soft tissue during flexion and extension. The stab incision at the lateral femoral epicondyle is potentially more significant since this has to pass through the tensor fascia latae and rectus femoris and thus could serve as a source of impingement of the soft tissue structures. However, bone pin insertion sites and stab incision direction were chosen carefully to minimize the damage and disruption on the skin, muscles, or ligaments.

The current study reported 3-Dimensional equine stifle passive kinematics in fully intact limbs. The findings share similarities with previously reported kinematics for humans, sheep, and dogs during passive joint motion, suggesting similar roles of the passive structures and overall joint function, despite obvious differences in the anatomic shapes, such as posterior tibial slopes, and joint loads. This suggests it may be possible to translate some of what has been learned in humans and dogs, such as mUKA, to treat equine stifle OA. The demonstrated method could also be used to investigate equine stifle kinematics in vivo as it has been reported to be feasible in producing accurate joint kinematics in other species. Future studies need to characterize in vivo equine stifle kinematics during different modes of gait and be compared with the current reported data.

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References


