Degeneration and ultimate rupture of the canine CrCL leads to stifle joint instability because the CrCL limits hyperextension, internal rotation, and cranial translation of the tibia relative to the femur.\(^1\),\(^2\) Rupture of the CrCL often is the result of degeneration over time; leads to inflammation, pain, lameness, and osteoarthritis; and is one of the most commonly diagnosed orthopedic conditions with a prevalence of 2.55\% across all breeds.\(^1\),\(^2\),\(^4\)\(^8\) Management costs of CrCL deficiency exceed $1 billion annually in the United States,\(^9\) and surgical stabilization is the standard of care for large dogs.\(^10\)

Several corrective intra-articular, extra-articular, and proximal tibial osteotomy reconstruction techniques have been proposed to stabilize CrCL-deficient stifles.\(^11\)\(^\text{-}^14\) Although there is evidence to suggest superiority of functional outcomes for osteotomy techniques,\(^11\),\(^15\) no technique has provided long-term prevention of pain or osteoarthritis progression.\(^16\)\(^\text{-}^20\) Therefore, an improved understanding of stifle joint biomechanical outcomes for nonsurgical management techniques is warranted.

Veterinary orthotic and prosthetic devices are being developed for animals with neurologic, orthopedic, and other pathological impairments.\(^21\)\(^\text{-}^27\) An orthotic device can be prescribed to provide a range of functions, including joint rest, immobilization, protection, control, movement assistance, movement prevention, and movement correction.\(^21\),\(^28\) Furthermore, orthoses may prevent the need for amputation,

**Biomechanics of an orthosis-managed cranial cruciate ligament–deficient canine stifle joint predicted by use of a computer model**

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**OBJECTIVE**
To evaluate effects of an orthosis on biomechanics of a cranial cruciate ligament (CrCL)–deficient canine stifle joint by use of a 3-D quasistatic rigid-body pelvic limb computer model simulating the stance phase of gait and to investigate influences of orthosis hinge stiffness (durometer).

**SAMPLE**
A previously developed computer simulation model for a healthy 33-kg 5-year-old neutered Golden Retriever.

**PROCEDURES**
A custom stifle joint orthosis was implemented in the CrCL-deficient pelvic limb computer simulation model. Ligament loads, relative tibial translation, and relative tibial rotation in the orthosis-stabilized stifle joint (baseline scenario; high-durometer hinge) were determined and compared with values for CrCL-intact and CrCL-deficient stifle joints. Sensitivity analysis was conducted to evaluate the influence of orthosis hinge stiffness on model outcome measures.

**RESULTS**
The orthosis decreased loads placed on the caudal cruciate and lateral collateral ligaments and increased load placed on the medial collateral ligament, compared with loads for the CrCL-intact stifle joint. Ligament loads were decreased in the orthosis-managed CrCL-deficient stifle joint, compared with loads for the CrCL-deficient stifle joint. Relative tibial translation and rotation decreased but were not eliminated after orthosis management. Increased orthosis hinge stiffness reduced tibial translation and rotation, whereas decreased hinge stiffness increased internal tibial rotation, compared with values for the baseline scenario.

**CONCLUSIONS AND CLINICAL RELEVANCE**
Stifle joint biomechanics were improved following orthosis implementation, compared with biomechanics of the CrCL-deficient stifle joint. Orthosis hinge stiffness influenced stifle joint biomechanics. An orthosis may be a viable option to stabilize a CrCL-deficient canine stifle joint. (Am J Vet Res 2017;78:27–35)

**ABBREVIATIONS**
CaCL  Caudal cruciate ligament
CrCL  Cranial cruciate ligament
LCL  Lateral collateral ligament
MCL  Medial collateral ligament
RTR  Relative tibial rotation
RTT  Relative tibial translation
offer an alternative to surgical intervention, and help restore an active lifestyle to maintain a high quality of life.\textsuperscript{21,24,25,29} Patients that are poor candidates for anesthesia, have important comorbidities, are of advanced age, or have owners who lack the financial means for costly surgeries may be candidates appropriately suited for an orthosis.\textsuperscript{25}

Orthosis efficacy is dependent on a customized design to restore function without causing trauma to tissues, and proper training of owners in orthosis use is required to improve the likelihood of compliance and patient acclimation to the device.\textsuperscript{28} Positive outcomes following implementation of an orthosis have been reported for dogs with sciatic neuropathy,\textsuperscript{24} strain of the tendinous portion of the gastrocnemius muscle,\textsuperscript{22} wounds on the pads of the feet,\textsuperscript{25} carpal ligament instability,\textsuperscript{27} and various other thoracic and pelvic limb pathological conditions.\textsuperscript{20,28} In some cases, the orthosis may provide limb or joint support during healing or rehabilitation or in association with concurrent treatments whereby use of the orthosis is discontinued following sufficient biomechanical improvement.\textsuperscript{22,25} However, long-term recovery of function in a CrCL-deficient stifle joint is not likely with orthosis use alone; therefore, physical rehabilitation and gait retraining are required,\textsuperscript{28} even though a stifle joint orthosis is often worn for most daily activities.

Similar to the situation for injury recovery in humans, rehabilitation has become an increasingly prominent aspect of veterinary care during the past decade.\textsuperscript{25} An improved understanding of quadrupedal motion and biomechanics has led to more sophisticatedorthoses with advanced designs and materials that can be custom fabricated for each patient to address specific needs.\textsuperscript{25,20,28} Typically, the orthosis allows limited functional movement, rather than complete immobilization, to improve healing while preventing further injury.\textsuperscript{27} The mechanical principles in orthotic management of CrCL-deficient canine stifle joints are similar to 4-point force application systems used in anterior cruciate ligament-deficient knees of humans.\textsuperscript{30} A typical canine stifle joint orthosis allows controlled articulation through cranial constraints at the quadriceps muscles, tibial tuberosity, and distal aspect of the tibia and caudal constraints at the semimembranosus and semitendinosus muscles, gastrocnemius muscle, and calcaneal tendon.\textsuperscript{26} Proper conformation between the orthosis and pelvic limb aligns the mechanical hinge center of rotation of the orthosis with the stifle joint center of rotation and limits internal tibial rotation and cranial tibial translation.\textsuperscript{26} However, no study has been conducted to quantify the stabilizing effect of an orthosis in a CrCL-deficient stifle joint.

Canine pelvic limb biomechanics have been investigated by use of computer simulation models of intact and CrCL-deficient stifle joints during the stance phase of gait.\textsuperscript{31,32} These models predicted increased CaCl loads, cranial tibial translation, and internal tibial rotation in CrCL-deficient stifle joints, which reinforced findings from in vivo and in vitro studies.\textsuperscript{2,3,35,34} Furthermore, computer simulation models of the canine pelvic limb are appropriately suited to investigate the influence of stifle joint biomechanical and anatomic parameters as well as effects of surgical interventions and other clinical scenarios.\textsuperscript{35,35–39} Computer models have predicted that stifle joint biomechanics are sensitive to ligament prestrain, tibial plateau angle, and femoral condyle size,\textsuperscript{31,35,39} and tibial plateau leveling osteotomy and tibial tuberosity advancement can affect stifle joint ligament loads and kinematics.\textsuperscript{36,37,39} A computer model of the canine pelvic limb is appropriately suited for use in investigating biomechanics of a CrCL-deficient stifle joint managed with an orthosis. Furthermore, the influence of orthosis design parameters can be evaluated with parametric sensitivity analysis.

The objectives of the study reported here were to implement a custom-fit canine stifle joint orthosis in a previously developed, anatomically accurate quasistatic 3-D computer model of a canine pelvic limb simulating the stance phase of gait; to evaluate ligament loading patterns, tibial translation, and tibial rotation in a CrCL-deficient stifle managed with an orthosis; and to investigate the influence of orthosis hinge stiffness (durometer) on stifle joint biomechanics. We hypothesized that a stifle joint orthosis would reduce cranial tibial translation and internal tibial rotation, compared with results for the CrCL-deficient stifle joint, and return ligament loads to those in a CrCL-intact stifle joint. Additionally, we hypothesized that orthosis hinge stiffness would affect stifle joint biomechanics.

**Materials and Methods**

**Canine pelvic limb computer model**

A 3-D quasi-static computer model of a canine pelvic limb previously developed on the basis of a 5-year-old 33-kg neutered Golden Retriever with no known orthopedic or neurologic disorders\textsuperscript{31} was used in the present study. This model has been used in other studies.\textsuperscript{31,35–38,40} The model was developed by the use of computer-aided design and modeling software\textsuperscript{4} and was evaluated by use of rigid-body motion simulation software.\textsuperscript{31,b} Model geometry was based on CT data, and motion capture kinematics and force platform kinetics were collected during the stance phase while the dog was walked on a leash in a straight line.\textsuperscript{31} Pelvic limb joint reactions were determined by use of inverse dynamics analysis, whereby stifle joint reaction moments were used to calculate forces in pelvic limb muscles crossing the stifle joint through the minimization of maximal muscle stress optimization strategy.\textsuperscript{31} Each ligament was represented as a tension-only element that carried load when stretched beyond its neutral length. Contact elements were included to prevent penetration between the femur and menisci, femur and tibia, and femur and patella.\textsuperscript{31}
Canine pelvic limb computer model with an implemented orthosis

A custom stifle joint orthosis was fabricated for the left pelvic limb of the dog represented in the 3-D computer model. The orthosis was based on a mold developed with a fiberglass impression of the thigh, crus, and tarsus while the dog was in a standing position. Thigh and crus cuffs were developed from the mold by use of thermoformed plastic. The orthosis was equipped with 2 hinges between the thigh and crus cuffs (one on the medial aspect and the other on the lateral aspect) that were free-motion composite orthotic joints featuring a high-strength inner core for smooth articulation. Foam padding and restraint straps were included in the orthosis to improve comfort and maintain proper alignment on the pelvic limb. Appropriate fit of the fabricated orthosis was verified on the dog before implementation into the computer model.

A 3-D scan of the custom stifle joint orthosis (with straps of the orthosis removed) was used to generate an orthosis point cloud of the thigh and crus cuffs. The orthosis point cloud, which is a set of coordinates in 3-D space representing geometry of the region scanned, was imported into modeling software, and converted to separate 3-D solid models of the thigh cuff and crus cuff. Furthermore, solid models of the thigh and crus soft tissues were developed from the canine pelvic limb model (Figure 1). Soft tissues were used to orient the stifle joint orthosis for each discrete interval of stance and to establish contact pairs between the stifle joint orthosis and canine pelvic limb to resist interpenetration by use of the following equation:

\[ F_n = (k g) + (d g / d t) f(g, c_{max}, d_{max}) \]

where \( F_n \) is the contact force, \( k \) is the stiffness, \( g \) is the contact penetration during simulation, \( c \) is the elastic component exponent, \( d g / d t \) is the instantaneous rate of change of \( g \) with respect to time, and \( f(g, c_{max}, d_{max}) \) is a contact penetration step function whereby \( c_{max} \) is the maximum damping and \( d_{max} \) is the penetration at which \( c_{max} \) occurs. For the contact equation, constants between the orthosis and canine pelvic limb were as follows: \( k = 1,150 \) N/mm, \( e = 2 \), \( c_{max} = 0.6 \) N-s/mm, and \( d_{max} = 0.1 \) mm. Furthermore, the thigh and crus soft tissues established contours for 4 restraint straps of the stifle joint orthosis (caudal thigh strap, caudal crus strap, caudal tarsus strap, and cranial crus strap). Each strap was represented as a straight-line, tension-only element connected in series contoured to soft tissue that applied load when stretched beyond each strap’s neutral length. For each strap, the first strap element was attached to the medial aspect of the orthosis, and the second strap element was attached to the lateral aspect of the orthosis; the 2 strap elements shared a fixed endpoint attached to the soft tissue geometry that represented wrapping around the soft tissue of the canine pelvic limb. Strap neutral length was determined at midstance.

Orthosis hinges were modeled as composite tension, compression, bending, and torsion elements by use of linear and torsional springs (Figure 1). Material properties were determined by use of a material testing machine for each orthosis strap and 2 unique hinges with differing stiffness (high-durometer hinge and low-durometer hinge). Hinges were tested in tension, compression, bending, and torsion, and orthosis straps were tested in tension. Additionally, the friction coefficient between the soft tissue and orthosis was set to 0.20. The friction coefficient was determined by measuring the force required to slide the orthosis foam padding with a known applied weight along a fur-covered surface. Finally, motion capture kinematics, ground reaction kinetics, and optimized muscle forces used in the previously developed model were applied to the canine pelvic limb model with the incorporated stifle joint orthosis.

Model simulation

The canine pelvic limb model simulated the stance phase of gait to determine CrCL, LCL, and MCL loads, tibial translation, and tibial rotation at discrete (10%) intervals. Relative tibial translation between the CrCL-intact stifle joint and CrCL-deficient stifle joint was defined as follows:

\[ RTT_{D/I} = (FT_{intact}) - (FT_{deficient}) \]

where \( FT \) is the fixed point tibial translation (distance between the tibial tuberosity position relative to a fixed point on the femur along the cranio-caudal axis), with deficient and intact indicating the stifle joint scenarios. Relative tibial translation between the CrCL-intact stifle joint and orthosis-managed stifle joint was defined as follows:

\[ RTT_{O/I} = (FT_{orthosis}) - (FT_{intact}) \]

where orthosis indicated the orthosis-managed stifle joint. Relative tibial rotation between the CrCL-intact stifle joint and CrCL-deficient stifle joint was defined as follows:

\[ RTR_{D/I} = (R_{deficient}) - (R_{intact}) \]

Relative tibial rotation between the CrCL-intact stifle joint and orthosis-managed stifle joint was defined as follows:

\[ RTR_{O/I} = (R_{orthosis}) - (R_{intact}) \]

where \( R \) is the internal-external rotation as defined elsewhere.

Sensitivity analysis for orthosis hinge stiffness

Sensitivity analysis was conducted by altering orthosis hinge stiffness in the computer simulation model. Influence of hinge stiffness on peak model outcomes was compared. Variation of hinge stiffness included simultaneous changes to tension stiffness,
compression stiffness, bending stiffness, and torsion stiffness to evaluate 5 scenarios (from least stiff to most stiff): 50% below stiffness of hinge B,\(^6\) stiffness of hinge B, stiffness midway between that of hinge B and that of hinge A,\(^7\) stiffness of hinge A (baseline scenario), and 50% above stiffness of hinge A.

**Data analysis**—Outcome measures for the orthosis-managed stifle joint were compared with outcome measures for the CrCL-intact and CrCL-deficient stifle joints by use of the peak value test.\(^6\)\(^3\) Statistical analyses with an ANOVA or similar tests were not performed because model-predicted outcomes were single values without variability. Differences between peak outcome measures for various models were considered significant if the absolute value was > 20% or if peak values were detected at stance phase intervals > 10%. Outcome measure sensitivity to hinge stiffness was assessed by use of a sensitivity index as follows: percentage change in outcome measure divided by percentage change in input parameter (ie, a 5% increase in outcome measure attributable to a 10% increase in input parameter represented a sensitivity index of 0.5).\(^6\)\(^3\) Higher absolute values of the sensitivity index indicate that the outcome measure is more sensitive to changes in the altered input parameter. Variation of hinge stiffness included simultaneous variation of multiple characteristics (ie, tension stiffness, compression stiffness, torsion stiffness, and bending stiffness) at different percentages because material properties did not differ equally between the 2 hinges evaluated. Therefore, change in a composite input parameter was defined as the mean percentage change in all input characteristics relative to the baseline scenario.

**Results**

Tension stiffness, compression stiffness, bending stiffness, and torsion stiffness were determined for each orthosis hinge, and tension stiffness was determined for each orthosis strap (Table 1). Ligament loads (Figure 2) and tibial translation and rotation (Figure 3) for the CrCL-deficient stifle joint managed with an orthosis were determined throughout the stance phase. Orthosis-managed stifle joint ligament loads exceeded the peak value test criteria, compared with loads for the CrCL-intact and CrCL-deficient stifle joints, which indicated a significant difference between the scenarios (Table 2). The

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**Table 1**—Orthosis component stiffness (durometer) determined through materials testing.

<table>
<thead>
<tr>
<th>Component</th>
<th>Tension (N/mm)</th>
<th>Compression (N/mm)</th>
<th>Bending (N-mm/°)</th>
<th>Torsion (N-mm/°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hinge A</td>
<td>181</td>
<td>67</td>
<td>50</td>
<td>6.1</td>
</tr>
<tr>
<td>Hinge B</td>
<td>96</td>
<td>21</td>
<td>25</td>
<td>2.5</td>
</tr>
<tr>
<td>Caudal thigh strap</td>
<td>94</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
<tr>
<td>Caudal crus strap</td>
<td>94</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
<tr>
<td>Caudal tarsus strap</td>
<td>3</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
<tr>
<td>Cranial crus strap</td>
<td>141</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
</tbody>
</table>

Hinge A was a high-durometer hinge,\(^6\) and hinge B was a low-durometer hinge.\(^4\) NA = Not applicable.

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CaCL and LCL loads were reduced, whereas MCL load increased, in the orthosis-managed stifle joint, compared with loads for the CrCL-intact stifle joint. All ligament loads were reduced in the orthosis-managed stifle joint, compared with loads for the CrCL-deficient stifle joint. Cranial RTT \( O/I \) was evident throughout the stance phase and peaked at 1.9 mm at 10% of the stance phase (Table 3). The RTR \( D/I \) peaked at 1.8° of internal rotation at 60% of the stance phase. When RTT \( D/I \) was compared to RTT \( O/I \) and RTR \( D/I \) was compared to RTR \( O/I \), the peak value test criteria were exceeded, which indicated significant differences between the modeled scenarios because peak cranial RTT \( D/I \) was 20.2 mm and peak internal RTR \( D/I \) was 4.6°.

For the sensitivity analysis, ligament loads (Figure 4) and RTT \( O/I \) and RTR \( O/I \) (Figure 5) throughout the stance phase were compared for each scenario of hinge stiffness. The minimum CaCL load occurred for the baseline scenario. The LCL load was similar for each scenario (ie, hinge stiffness had negligible influence on LCL load), whereas MCL load slightly decreased as hinge stiffness increased. Peak RTT \( O/I \) was similar to baseline values for all scenarios, except for the 50% increase in stiffness, which led to a decrease in cranial tibial translation. The RTR \( O/I \) decreased with increasing hinge stiffness. Each outcome measure was sensitive to orthosis hinge stiffness, as indicated by sensitivity indices (Table 4).

**Discussion**

In the present study, CaCL and LCL loads were reduced 38% and 53%, respectively, throughout the stance phase in the orthosis-managed stifle joint, compared
with loads for the CrCL-intact stifle joint. The MCL loads increased at midstance from a ligament load of 10% of body weight in the CrCL-intact stifle joint to a ligament load of 15% of body weight in the orthosis-managed stifle joint, an increase of 50%. Therefore, the peak value test criteria were exceeded, which indicated that orthosis-managed stifle joint ligament loads were significantly different from those of the intact stifle joint. Peak CaCL, LCL, and MCL loads were reduced 90%, 93%, and 59%, respectively, for the orthosis-managed stifle joint, compared with loads for the CrCL-deficient stifle joint, thereby exceeding the peak value test criteria, which indicated significant differences. Little is known regarding in vivo CaCL, LCL, and MCL loading in the orthosis-managed stifle joint, and to the authors’ knowledge, no in vitro study has been conducted to investigate biomechanics of orthosis-managed stifle joints. However, findings for the computer model suggested improvement in stifle joint stability through decreased ligament loading and better management of kinematics in abnormal stifle joints. Peak RTT \( D/I \) and RTR \( D/I \) decreased 91% and 61%, respectively, compared with RTT \( D/I \) and RTR \( D/I \), thereby exceeding the peak value test criteria, which indicated significant differences between the scenarios. Therefore, the orthosis improved stability in the CrCL-deficient stifle joint, partially restored normal biomechanics, and potentially reduced the likelihood for injury of the CaCL, LCL, and MCL in a CrCL-deficient stifle joint because ligament loads were substantially less than those for the CrCL-deficient state in the orthosis-managed stifle joint. These findings are important given that there is an increased likelihood of CaCL injury attributable to morphological changes and repetitive microtrauma in CrCL-deficient stifle joints.
Overall, evaluation of ligament loading and stifle joint kinematics through computer modeling in the present study suggested that an orthosis-managed stifle joint offered a biomechanically sound approach for management of a CrCL-deficient stifle joint.

The baseline hinge stiffness scenario in the model corresponded to a flexible composite hinge used clinically in human and veterinary orthoses and was included in the orthosis fabricated for the dog used to develop the model for this study. However, other hinge stiffness values are available and may be used depending on the orthosis application and the patient’s needs. Therefore, we chose to evaluate additional hinge stiffness values that corresponded to a lower-stiffness, veterinary-specific composite hinge as well as variants of both hinges to investigate the influence of hinge material properties on biomechanical efficacy of a stifle joint orthosis. Stiffness values were on average 56% less for the lower-stiffness, veterinary-specific hinge (hinge B), compared with stiffness values for the baseline scenario (hinge A). The computer model predicted that changes in hinge stiffness influenced the stabilizing effect of the orthosis. Peak absolute value for sensitivity indices across all outcomes ranged from 0.46 to 1.09, and at least 1 biomechanical outcome was most sensitive to each hinge stiffness evaluated in the analysis. The RTR_{O/I} was most sensitive (-0.70 to -1.09) to changes in hinge stiffness, whereas RTT_{O/I} was least sensitive (-0.03 to -0.46) to changes in hinge stiffness. As hinge stiffness increased (ie, stiffer hinge), internal tibial rotation decreased. Therefore, internal rotation can be controlled by modifying hinge stiffness, but greater control of tibial translation may require modification of other orthosis design parameters or the use of hinges with material properties that differ from those evaluated here. Increasing hinge stiffness decreases flexibility of an orthosis, but decreased flexibility would require greater moment generation about the stifle joint, which would require greater muscle strength to achieve full range of motion of the joint. This may be of concern in a dog with CrCL deficiency because muscle atrophy may decrease the strength of muscles crossing the stifle joint. Therefore, it is important to select the appropriate orthosis hinge that stabilizes but does not overconstrain the stifle joint. Sensitivity analysis in the present study investigated variation of properties in one type of orthosis hinge design. Investigation of other hinge types, including monocentric and polycentric hinges that constrain orthosis degrees of freedom, is warranted.

Findings from the present study should be considered with respect to the following study limitations. Soft tissue geometry was based on CT data, which were obtained while the dog was in lateral recumbency. Soft tissue contours may differ during weight-bearing postures, compared with during lat-

### Table 4—Sensitivity indices for hinge stiffness variation.

<table>
<thead>
<tr>
<th>Hinge stiffness scenario</th>
<th>CaCL load</th>
<th>LCL load</th>
<th>MCL load</th>
<th>RTT_{O/I}</th>
<th>RTR_{O/I}</th>
</tr>
</thead>
<tbody>
<tr>
<td>50% below hinge B</td>
<td>-78</td>
<td>-0.47</td>
<td>-0.01</td>
<td>-0.50*</td>
<td>-0.03</td>
</tr>
<tr>
<td>Hinge B</td>
<td>-56</td>
<td>-0.86*</td>
<td>0.10</td>
<td>-0.44</td>
<td>-0.05</td>
</tr>
<tr>
<td>Midway</td>
<td>-28</td>
<td>-0.05</td>
<td>-0.67*</td>
<td>-0.34</td>
<td>-0.06</td>
</tr>
<tr>
<td>Hinge A</td>
<td>0</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
<tr>
<td>50% above hinge A</td>
<td>50</td>
<td>0.39</td>
<td>0.32</td>
<td>-0.20</td>
<td>-0.46*</td>
</tr>
</tbody>
</table>

Hinge stiffness variation included simultaneous variation of multiple characteristics (ie, tension stiffness, torsion stiffness, and bending stiffness); thus, a composite stiffness change was defined as the mean percentage change in all stiffness characteristics relative to that for hinge A (baseline scenario). Midway represented the stiffness halfway between that of hinge A and that of hinge B. *Peak value for the outcome.

See Table 1 for remainder of key.
eral recumbency, as well as across the stance phase, and these differences were not represented in this study. These differences can influence orthosis fit as well as the contours of orthosis straps around the soft tissue in the model. The orthosis cuffs contained foam adhered to a plastic shell. However, the orthosis cuff models were treated as a single material. Contact pairs between the orthosis cuffs and soft tissue of the pelvic limb were considered nearly rigid to resist penetration between the orthosis and canine pelvic limb. Deformation of the foam and soft tissue may influence orthosis alignment and fit to the pelvic limb in vivo, and these small deformations were simplified in the model.

The orthosis straps were modeled as straight-line elements that attached to the orthosis and pelvic limb. This approach constrained the orthosis to the pelvic limb and prevented proximodistal translation of the strap relative to the pelvic limb. Additionally, curvature of the straps around the pelvic limb was simplified by use of 2 straight-line elements per strap. Furthermore, strap slack is possible when owners improperly adjust strap tension. Straps that are too tight or too slack were not investigated in the present study. Strap tension, the number of straps, and the locations of straps should be assessed in future parametric analyses.

Findings were reported for a model developed on the basis of only one dog, and orthosis efficacy may differ in other dogs as a result of individual characteristics such as skin movement, coat thickness, and limb conformation. Furthermore, biomechanical and morphological differences, including tibial plateau angle, ligament tension, and femoral condyle shape, may affect stifle joint biomechanics, but the influence of these parameters in orthosis-managed stifle joints is unknown.

The menisci were represented in the present model as bodies with constant geometry determined from CT data obtained from a healthy dog. In vivo deformation of meniscus shape may affect stifle joint biomechanics, including ligament loading. Furthermore, meniscus integrity may degrade following CrCL deficiency, and this degradation was not represented in the model. The computer model simulation was conducted at 1 time point and did not account for longitudinal changes in in vivo conditions (eg, progression of osteoarthritis in a CrCL-deficient stifle joint or pelvic limb muscle atrophy). Therefore, quantification of in vivo stability of an orthosis-managed stifle joint is needed to support results from the present study. Furthermore, kinematics and kinetics applied to the model were determined for a healthy dog without an orthosis. Kinematics and kinetics may differ for a dog with an orthosis. Finally, the computer model was used to evaluate the stance phase of walking gait, which is susceptible to dynamic stifle joint instability attributable to active weight bearing. Other biomechanical scenarios (eg, swing phase of the gait or during sitting, turning, trotting, running, and jumping) were not evaluated.

In the present study, biomechanics in a CrCL-deficient stifle joint were improved following implementation of a stifle joint orthosis. Orthosis hinge stiffness influenced stifle joint biomechanics.

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Contents of this manuscript are solely the responsibility of the authors and do not necessarily represent the views of the American Kennel Club Canine Health Foundation.

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Footnotes

b. SolidWorks Motion, version 2012, Structural Research and Analysis Corp, Santa Monica, Calif.
c. Cranial cruciate orthosis, OrthoPets V-OP Veterinary Clinic, Denver, Colo.
d. Desktop 3D scanner, NextEngine Inc, Santa Monica, Calif.
e. MTESTQuattro, Admet Inc, Norwood, Mass.
f. Tamarack flexure joint, model 740-L, Tamarack Habilitation Technologies Inc, Blaine, Minn.
g. Tamarack flexure joint, model VET-L-65, Tamarack Habilitation Technologies Inc, Blaine, Minn.

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