An intact cranial cruciate ligament (CrCL) is vital to the craniocaudal and rotational stability of the stifle. The CrCL resists both cranial tibial translation and internal tibial rotation. The optimal surgical treatment for the CrCL-deficient stifle should provide immediate stabilization of the stifle while allowing normal joint contact mechanics and movement in all anatomic planes. Normal canine stifle movement during walking gait consists of flexion and extension ranging from 100–160°, internal and external tibial rotation up to 10°, and varus and valgus tibial movements up to 10°. During the stance phase of walking gait, stifle flexion–extension ranges from 120–145°. Procedures for stabilization of the CrCL-deficient stifle include extracapsular suture prostheses using monofilament or multifilament braided materials or procedures that modify bone morphology, such as the tibial tuberosity advancement (TTA) or tibial plateau leveling osteotomy (TPLO). Extracapsular suture temporarily stabilizes the CrCL-deficient joint until sufficient periarticular fibrosis provides some degree of stability. Both TPLO and TTA are purported to provide dynamic stability to the joint during the weight-bearing phase of gait. However, they do not prevent abnormal tibial thrust during all phases of gait, they alter joint contact mechanics, and can reduce range of motion postoperative. Progressive osteoarthritis is a consequence for all stifles regardless of the surgical procedure performed and although these procedures improve stifle stability, all are suboptimal. The Simitri Stable in Stride™ (New Generation Devices, Glen Rock, NJ) extracapsular articulating implant (EAI) was developed as a possible alternative to stabilize the canine CrCL- or caudal cruciate (CaCL)-deficient stifle. To be effective, the EAI must tolerate in vivo joint forces during gait while preventing abnormal movement of the tibia relative to the femur. Computer models have been used to investigate canine pelvic limb biomechanics and in silico studies complement in vivo and in vitro studies. Furthermore, canine pelvic limb computer simulation models have been used to investigate ligament-loading patterns and stifle kinematics in the CrCL-intact stifle, CrCL-deficient stifle, and TPLO-managed stifle.
during the stance phase of gait. These studies incorporated subject-specific anatomically accurate bone geometry, kinematics, and kinetics, along with active pelvic limb muscle forces. The sensitivity of biomechanical outcomes to variation of anatomic and surgical parameters has also been predicted using computer models.6,17,18,21

The goals of this study were to (1) implement the EAI into our previously developed quasi-static canine pelvic limb 3-dimensional computer simulation model, and (2) evaluate stifle ligament loading patterns, tibial translation, and tibial rotation during the stance phase of gait. We hypothesized that the EAI-managed CrCL-deficient stifle would decrease peak stifle ligament loading, decrease peak relative tibial translation, and decrease peak relative tibial rotation compared to the CrCL-deficient stifle across the stance phase of gait.

MATERIALS AND METHODS

Canine Pelvic Limb Computer Model

A canine pelvic limb computer model was previously developed from the left pelvic limb of a 5-year-old neutered male golden retriever (33 kg) with no known orthopedic or neurologic disorders (Fig 1). The 3D quasi-static model was developed using computer-aided design modeling software (SolidWorks v. 2012, SolidWorks Corp., Concord, MA) and the stance phase of gait was simulated using rigid body motion simulation software (SolidWorks Motion v. 2012, Structural Research and Analysis Corp, Santa Monica, CA).6 The dog was leash-walked in a straight line, and motion capture kinematic and force platform kinetic data were collected during the stance phase. An inverse dynamics analysis was conducted to determine pelvic limb joint reactions. Forces in 22 muscles crossing the stifle were determined throughout stance using the minimization of maximal muscle stress strategy and the stifle reaction moments determined from inverse dynamics analysis.6 Each muscle was represented as a linear force vector acting along the muscle line of action.6 Stifle ligaments were represented as tension-only elements that were loaded when stretched beyond their neutral length.6 Finally, contact elements between the femur and menisci, the femur and tibia, and the femur and patellar prevented penetration between stifle geometry.6

Canine Pelvic Limb Model With Implemented Extracapsular Articulating Implant

Clinical implementation of the EAI has been described previously.22 Therefore, only the EAI biomechanical function pertinent to model implementation is described here. The EAI consists of 3 components implanted on the medial aspect of the stifle: (1) femoral component, (2) travel channel insert, and (3) tibial component (Fig 1). The travel channel insert is compression fit into the proximal end of the tibial

Figure 1  (A) Canine pelvic limb model with muscle lines of action crossing the stifle. (B) Extracapsular articulating implant components positioned relative to the stifle, including the (1) femoral component, (2) travel channel insert, and (3) tibial component.
component. Engagement of the femoral ball component in the travel channel allows for sliding articulation between the 2 components.

The EAI was implemented in the canine pelvic limb model with the stifle in full extension (160° sagittal plane angle) using three-dimensional modeling software (SolidWorks v. 2012, SolidWorks Corp.). Model components included the femoral component, the tibial component, and the travel channel insert. The femoral and tibial components were contoured to match anatomical geometry and fixed to the distal femur and proximal tibia. The femoral component long axis was oriented along the femur long axis, and the femoral ball was placed caudal to the midpoint of Blumenstaat’s line so that the distance in the sagittal plane from the femoral ball center to the caudal and distal margins of the femoral condyles was 3.8 mm and 3.2 mm, respectively. The travel channel insert was fixed into the proximal opening of the tibial component overlying the femoral ball such that the ball was centered in the channel and the proximal extent of the tibial component was parallel to the distal extent of the femoral component. Finally, the tibial component long axis was oriented along the tibia long axis. Each component was modeled as a rigid body. Motion capture kinematics, ground reaction forces, and optimized muscle forces from our previously developed model were implemented in the canine pelvic limb model modified with the EAI.

**Stance Phase of Gait Simulation**

Rigid body motion software (SolidWorks Motion v. 2012, Structural Research and Analysis Corp.) was used to simulate the canine pelvic limb model during the stance phase of gait in 10% discrete intervals corresponding to stifle flexion-extension angles ranging from 120–135°. The positions of the femoral and tibial components with its associated travel channel were maintained for each stance phase. During simulation, the femoral ball and tibial travel channel interacted through contact elements to allow motion between the 2 components while preventing interpenetration using

\[ F_n = k g e + \left( \frac{d g}{dt} \right) f(g, c_{\text{max}}, d_{\text{max}}) \]  

where \( F_n \) is the contact force, \( k \) is the stiffness, \( e \) is the elastic component exponent, \( d g/dt \) is the instantaneous rate of change of \( g \) with respect to time, and \( f(g, c_{\text{max}}, d_{\text{max}}) \) is a contact penetration step function, whereby \( c_{\text{max}} \) is the maximum damping and \( d_{\text{max}} \) is the penetration at which \( c_{\text{max}} \) occurs. Contact equation constants for the EAI used in the study were as follows: \( k = 15,000 \text{ N/mm}, \ e = 1.5, \ c_{\text{max}} = 50 \text{ N-s/mm}, \) and \( d_{\text{max}} = 0.5 \text{ mm} \) to represent contact between the steel ball and plastic travel channel. The static and dynamic friction coefficients were set to 0.05 and 0.08, respectively.

Model outcomes included CaCL, lateral collateral ligament, medial collateral ligament and patellar ligament loads, and tibial translation and rotation relative to the femur. The relative tibial translation between the CrCL-intact and the CrCL-deficient stifle was defined as

\[ \text{RTT}_{\text{D/1}} = (\text{FT}_{\text{deficient}}) - (\text{FT}_{\text{intact}}) \]

where \( \text{FT}_{\text{deficient}} \) represents fixed point tibial translation (i.e., the distance between the tibial tuberosity position relative to a fixed point on the femur in the cranio-caudal direction) for the CrCL-deficient stifle and \( \text{FT}_{\text{intact}} \) is the fixed point tibial translation for the CrCL-intact stifle. The relative tibial translation between the CrCL-intact and the EAI-managed stifle was defined as

\[ \text{RTT}_{\text{EAI/1}} = (\text{FT}_{\text{EAI}}) - (\text{FT}_{\text{intact}}) \]

where \( \text{FT}_{\text{EAI}} \) represents fixed point tibial translation for the EAI-managed stifle. Cranial relative tibial translation was defined as positive and caudal relative tibial translation was defined as negative.

The relative tibial rotation between the CrCL-intact and the CrCL-deficient stifle was defined as

\[ \text{RTR}_{\text{D/1}} = (\text{R}_{\text{deficient}}) - (\text{R}_{\text{intact}}) \]

where \( \text{R}_{\text{deficient}} \) is the internal–external rotation for the CrCL-deficient stifle and \( \text{R}_{\text{intact}} \) is the internal–external rotation for the CrCL-intact stifle. The relative tibial rotation between the CrCL-intact and the EAI-managed stifle was defined as

\[ \text{RTR}_{\text{EAI/1}} = (\text{R}_{\text{EAI}}) - (\text{R}_{\text{intact}}) \]

where \( \text{R}_{\text{EAI}} \) is the internal–external rotation for the EAI-managed stifle. Internal relative tibial rotation was defined as positive and external relative tibial rotation was defined as negative.

**Data Analysis**

Model outcomes were compared to previously published model-predicted CrCL-intact and CrCL-deficient stifle outcomes using the peak value test, the ratio between outcome average stance interval histories, correlation coefficient, and the standard deviation of residuals. Evaluation criteria for significant differences were as recommended by Pipkorn and Eriksson. Differences between various model-predicted peak outcomes were considered significant if the absolute difference was >20% or if peak values occurred at stance phase intervals that were more than 10% apart. The ratio between outcome average stance interval histories, \( \bar{s} \), was defined as

\[ \bar{s} = \frac{\bar{t}_i}{\bar{t}_j} \]

where, \( \bar{t}_i \) is the average of an outcome stance interval history for scenario i (e.g., CrCL-intact stifle, CrCL-deficient stifle, EAI-managed stifle), and \( \bar{t}_j \) is the average of the outcome stance interval history for scenario j. Outcome stance interval histories for the compared scenarios were
considered different if \( s < 0.8 \) or \( 1.2 < s \). The correlation coefficient was defined as

\[
\rho = \frac{\sum_{i=0}^{n} (f_i - \bar{f}_i)(f_j - \bar{f}_j)}{\sqrt{\sum_{i=0}^{n} (f_i - \bar{f}_i)^2 \sum_{j=0}^{n} (f_j - \bar{f}_j)^2}}
\]

where, \( f_i \) is the \( i \)th value of the outcome stance interval history for one scenario, \( f_j \) is the \( j \)th value of the outcome stance interval history for another scenario, and \( i \) and \( j \) range from 0 to \( n = 100 \), incremented by 10 to represent comparison of 10% stance intervals. Outcome stance interval histories for the compared scenarios were considered different for \( \rho < 0.8 \). The standard deviation of residuals, \( \sigma_e \), was defined as

\[
\sigma_e = \sqrt{\frac{n \sum_{i=0}^{n} (f_i - f_j)^2 - \sum_{i=0}^{n} (f_i - f_j)^2}{n(n-1)}}
\]

and the compared outcome stance interval histories were considered different when \( \sigma_e > 20\% \) of the outcome peak value for the comparative scenario. If an evaluation criterion was exceeded for 1 of the 4 comparative tests, then the 2 scenarios were deemed to be different for the outcome measure being evaluated. Statistical analyses were not performed because model-predicted outcomes at each discrete stance phase interval are single values without variability.

RESULTS

Ligament loads (Fig 2), and tibial translation and rotation (Figs 3 and 4) for the CrCL-deficient stifle managed with the EAI were determined throughout stance. The CaCL, lateral collateral ligament, and medial collateral ligament loads in the EAI-managed stifle were different from those for the CrCL-intact stifle (peak values, \( \rho \), \( \sigma_e \), and \( s \) evaluation criteria exceeded, Table 1). There was no difference between the patellar ligament loads in the CrCL-intact and EAI-managed stifle. Peak CaCL and medial collateral ligament loads in the EAI-managed stifle increased 148% and 62%.

Figure 2  Ligament load as percent body weight (BW) throughout stance in the CrCL-intact (solid line with black diamonds), CrCL-deficient (dotted line with white squares), and extracapsular articulating implant (EAI)-managed stifle (dashed line with gray triangles). (A) Caudal cruciate ligament (CaCL). (B) Lateral collateral ligament (LCL). (C) Medial collateral ligament (MCL). (D) Patellar ligament (PL). There were significant differences in peak CaCL, LCL, and MCL loads between the intact and EAI-managed stifle and the CrCL-deficient and EAI-managed stifle. There was no significant difference in peak PL loads between the CrCL-intact and EAI-managed stifle, or between the CrCL-deficient and EAI-managed stifle. (Note: Connecting lines between data points are for visual purposes and do not represent values obtained from simulation.)
respectively, and lateral collateral and patellar ligament loads decreased 16% and 1%, respectively, compared to the CrCL-intact stifle. The CaCL, lateral collateral ligament and medial collateral ligament loads in the EAI-managed stifle were different than those of the CrCL-deficient stifle (peak values, \( r, \sigma_e, \) and \( \bar{s} \) evaluation criteria exceeded, Table 1). No differences were found in patellar ligament loads loads between the EAI-managed stifle and the CrCL-deficient stifle. Peak CaCL, lateral collateral ligament, and medial collateral ligament loads in the EAI-managed stifle decreased 94%, 100%, and 62%, respectively, compared to the CrCL-deficient stifle.

Peak RTTEAI was 1.0 mm caudal (20% stance), representing a significant reduction (105%) compared to RTTD (peak values, \( r, \sigma_e, \) and \( \bar{s} \) evaluation criteria exceeded, Table 2). Tibial translation in the EAI-managed stifle was in a caudal direction compared to tibial translation in a cranial direction in the CrCL-deficient stifle. Peak RTREAI was significantly reduced compared to the RTRDI peak of 2.4° internal rotation at 60% stance (peak values, \( r, \sigma_e, \) and \( \bar{s} \) evaluation criteria exceeded, Table 2). This represented a reduction of 48% compared to RTRDI, the relative rotation between the CrCL-intact and CrCL-deficient stifle. Comparison of absolute tibial translation and rotation values showed peak absolute translation in the CrCL-intact, CrCL-deficient, and EAI-managed stifle was 1.2 mm (cranial, 50% stance), 21.4 mm (cranial, 50% stance), and 2.1 mm (cranial, 50% stance), respectively, and peak absolute rotation in the CrCL-intact, CrCL-deficient, and EAI-managed stifle was 8.9° (external, 60% stance), 7.9° (external, 60% stance), and 6.5° (external, 60% stance), respectively.

DISCUSSION

Computer simulation models have previously been used to investigate stifle biomechanics in the CrCL-intact, CrCL-deficient, and surgically managed CrCL-deficient stifles.6,17–21 In the present study, a novel EAI was applied to a previously

Figure 3  (A) Relative tibial translation throughout the stance phase between the CrCL-deficient and CrCL-intact stifle (RTTD, black diamonds) and between the extracapsular articulating implant (EAI)-managed and the CrCL-intact stifle (RTTEAI, white squares). A positive value corresponds to cranial translation, and a negative value corresponds to caudal translation. There was a significant difference between peak RTTD and peak RTTEAI. (B) Relative tibial rotation throughout the stance phase between the CrCL-deficient and CrCL-intact stifle (RTRD, black diamonds) and between the EAI-managed and CrCL-intact stifle (RTREAI, white squares). A positive value corresponds to internal rotation, and a negative value corresponds to external rotation. There was a significant difference between peak RTRD and peak RTREAI. (Note: Connecting lines between data points are for visual purposes and do not represent values obtained from simulation.)

Figure 4  (A) Absolute tibial translation throughout the stance phase in the CrCL-intact (black diamonds), CrCL-deficient (white squares), and extracapsular articulating implant (EAI)-managed stifle (black crosses). A positive value corresponds to cranial translation, and a negative value corresponds to caudal translation. (B) Absolute tibial rotation throughout the stance phase in the CrCL-intact (black diamonds), CrCL-deficient (white squares), and EAI-managed stifle (black crosses). A positive value corresponds to internal rotation, and a negative value corresponds to external rotation. (Note: Connecting lines between data points are for visual purposes and do not represent values obtained from simulation.)
developed computer simulation model of the canine pelvic limb to evaluate stifle ligament loads and tibial kinematics during the stance phase of gait. Significant decreases in peak CaCL, lateral collateral ligament, and medial collateral ligament loads were predicted in the EAI-managed stifle compared to the CrCL-deficient stifle. Peak relative tibial translation and rotation were also significantly reduced with implementation of the EAI in the CrCL-deficient stifle. Peak patellar ligament loads were not significantly different between the CrCL-intact, CrCL-deficient, and EAI-managed stifles.

Given that peak ligament loading occurs at mid-stance in the CrCL-deficient stifle, CaCL, lateral collateral ligament, medial collateral ligament, and patellar ligament loads at 50% stance were compared for the various stifle conditions. This is also consistent with in vitro studies which often evaluate changes after surgical procedures at mid-stance. At 50% stance, CaCL, lateral collateral ligament, medial collateral ligament, and patellar ligament loads in the EAI-managed stifle were significantly reduced (by 95%, 100%, 57%, and 58%, respectively) compared to the CrCL-deficient stifle, and were more similar to the CrCL-intact stifle. However, in the early stance phase, the model predicted an EAI-managed peak CaCL load of 72% BW at 10% stance, which exceeded the CaCL load in the CrCL-deficient and CrCL-intact stifles at that stance interval. Subsequent local CaCL load maxima occurred at 60% and 90% stance. These local maxima corresponded to local maxima in RTRD/I, suggesting that internal rotation may load the CaCL during the latter phase of stance. These findings are supported by an in

Table 1  Peak Caudal Cruciate Ligament (CaCL), Lateral Collateral Ligament (LCL), Medial Collateral Ligament (MCL), and Patellar Ligament (PL) Loads (% Body Weight, BW) Throughout Stance

<table>
<thead>
<tr>
<th>Ligament Load</th>
<th>Peak Load (% BW)</th>
<th>% △ From Intact</th>
<th>% △ From Deficient</th>
<th>Load at 50% Stance</th>
<th>% △ From Intact</th>
<th>% △ From Deficient</th>
<th>EAI/I vs D/I Intact</th>
<th>s</th>
<th>r</th>
<th>σa (% BW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CrCL-intact</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>---</td>
<td>---</td>
<td>----------</td>
</tr>
<tr>
<td>CaCL</td>
<td>29% BW (10%)</td>
<td>—</td>
<td>—</td>
<td>7% BW (50%)</td>
<td>—</td>
<td>—</td>
<td>84</td>
<td>0.67</td>
<td>0.48</td>
<td>18.6</td>
</tr>
<tr>
<td>LCL</td>
<td>15% BW (90%)</td>
<td>—</td>
<td>—</td>
<td>2% BW (50%)</td>
<td>—</td>
<td>—</td>
<td>96</td>
<td>1.48</td>
<td>0.57</td>
<td>4.4</td>
</tr>
<tr>
<td>MCL</td>
<td>10% BW (40%)</td>
<td>—</td>
<td>—</td>
<td>5% BW (50%)</td>
<td>—</td>
<td>—</td>
<td>86</td>
<td>0.60</td>
<td>0.45</td>
<td>4.5</td>
</tr>
<tr>
<td>PL</td>
<td>64% BW (90%)</td>
<td>—</td>
<td>—</td>
<td>9% BW (50%)</td>
<td>—</td>
<td>—</td>
<td>73</td>
<td>0.98</td>
<td>1.00</td>
<td>1.0</td>
</tr>
<tr>
<td>CrCL-deficient</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>---</td>
<td>---</td>
<td>----------</td>
</tr>
<tr>
<td>CaCL</td>
<td>183% BW (50%)†</td>
<td>531</td>
<td>—</td>
<td>183% BW† (50%)</td>
<td>2514</td>
<td>—</td>
<td>13.9</td>
<td>1.30</td>
<td>0.01</td>
<td>59.3</td>
</tr>
<tr>
<td>LCL</td>
<td>104% BW (50%)†</td>
<td>593</td>
<td>—</td>
<td>104% BW† (50%)</td>
<td>5100</td>
<td>—</td>
<td>4.45</td>
<td>0.19</td>
<td>31.5</td>
<td></td>
</tr>
<tr>
<td>MCL</td>
<td>37% BW (50%)†</td>
<td>270</td>
<td>—</td>
<td>37% BW† (50%)</td>
<td>640</td>
<td>—</td>
<td>1.20</td>
<td>0.83</td>
<td>7.3</td>
<td></td>
</tr>
<tr>
<td>PL</td>
<td>66% BW (90%)†</td>
<td>3</td>
<td>—</td>
<td>26% BW† (50%)</td>
<td>189</td>
<td>—</td>
<td>1.07</td>
<td>0.98</td>
<td>4.8</td>
<td></td>
</tr>
<tr>
<td>EAI-managed</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>---</td>
<td>---</td>
<td>----------</td>
</tr>
<tr>
<td>CaCL</td>
<td>72% BW (10%)††</td>
<td>148</td>
<td>—</td>
<td>10% BW†† (10%)</td>
<td>30</td>
<td>—</td>
<td>95</td>
<td>0.13</td>
<td>0.46</td>
<td>1.4</td>
</tr>
<tr>
<td>LCL</td>
<td>12% BW (20%)††</td>
<td>—</td>
<td>—</td>
<td>0% BW†† (20%)</td>
<td>—</td>
<td>—</td>
<td>100</td>
<td>1.45</td>
<td>0.19</td>
<td>7.3</td>
</tr>
<tr>
<td>MCL</td>
<td>16% BW (50%)††</td>
<td>62</td>
<td>—</td>
<td>16% BW†† (50%)</td>
<td>220</td>
<td>—</td>
<td>57</td>
<td>0.45</td>
<td>1.4</td>
<td></td>
</tr>
<tr>
<td>PL</td>
<td>63% BW (90%)††</td>
<td>—</td>
<td>—</td>
<td>11% BW†† (90%)</td>
<td>22</td>
<td>—</td>
<td>58</td>
<td>1.20</td>
<td>0.83</td>
<td>31.5</td>
</tr>
</tbody>
</table>

Table 2  Peak Relative Tibial Translation and Rotation Between Cranial Cruciate Ligament (CrCL)-Intact and CrCL-Deficient Stifle (D/I), and CrCL-Intact and Extracapsular Articulating Implant (EAI)-Managed Stifle (EAI/I)

<table>
<thead>
<tr>
<th>Outcome</th>
<th>Peak Value (Stance Interval, %)</th>
<th>% △ From D/I</th>
<th>% △ From EAI/I</th>
<th>50% Stance</th>
<th>EAI/I vs D/I</th>
</tr>
</thead>
<tbody>
<tr>
<td>RTD/I</td>
<td>20.2 mm (50)</td>
<td>—</td>
<td>—</td>
<td>20.2 mm</td>
<td>24.5†‡±3</td>
</tr>
<tr>
<td>RTTEAI</td>
<td>−1.0 mm (20)</td>
<td>−105</td>
<td>−96</td>
<td>0.9 mm‡</td>
<td>0.13†‡±3</td>
</tr>
<tr>
<td>RTD/EAI</td>
<td>4.6° (50)</td>
<td>—</td>
<td>—</td>
<td>4.6°</td>
<td>0.72†‡±3</td>
</tr>
<tr>
<td>RTP/EAI</td>
<td>2.4° (60)*</td>
<td>−48</td>
<td>−61</td>
<td>1.8°</td>
<td>0.46†‡±3</td>
</tr>
</tbody>
</table>

Values are also reported at 50% stance. The ratio of average stance interval histories (s), correlation coefficient (r), and standard deviation of residuals (σa) between the EAI-managed stifle and CrCL-intact and CrCL-deficient stifles are provided.

†Significant difference compared to the CrCL-intact stifle (peak value test criteria exceeded).
|
| †Significant difference compared to the CrCL-deficient stifle (peak value test criteria exceeded).

© Copyright 2016 by The American College of Veterinary Surgeons
vitro study that indicated transection of the CaCL led to an increase in passive stifle internal rotation in both flexion and extension. The computer simulation model predicted a peak medial collateral ligament load of 16% BW at 50% stance in the EAI-managed stifle which was slightly more than that of the CrCL-intact stifle at that stance interval. However, medial collateral ligament load subsequently decreased to more closely match the CrCL-intact stifle for the remainder of stance. The peak lateral collateral ligament load in the EAI-managed stifle occurred at 20% stance and was less than the peak lateral collateral ligament load in the CrCL-intact stifle, which occurred at 90% stance. This study predicted that the EAI-managed stifle decreased peak ligament loading compared to the CrCL-deficient stifle, which may reduce the possibility of remaining primary stifle support structure injury compared to the unmanaged deficient stifle. EAI-managed peak lateral collateral ligament and patellar ligament loads were predicted to be less than those in the CrCL-intact stifle. However, the EAI did not reduce predicted CaCL and medial collateral ligament peak loads to their natural state, and therefore may not entirely mitigate the possibility of injury to these ligaments given the higher than physiologic ligament loading. Additional longitudinal investigation of the EAI-managed stifle is needed to determine if these increased CaCL and MCL loads are clinically relevant.

Early stabilization and mobilization of the unstable stifle may improve clinical outcomes. For instance, medial meniscus injury is common after CrCL deficiency because tibial subluxation shifts femorotibial contact forces caudally leading to repetitive impingement of the meniscus caudal horn. Relative tibial translation and rotation significantly decreased from 20.2 mm (RTTD/I, cranial) and 4.6° (RTRD/I, internal rotation) to 1.0 mm (RTTEAI/I, caudal) and 2.4° (RTRTEAI/I, internal rotation), respectively, in the EAI-managed stifle. This decrease in peak tibial kinematics relative to the femur suggests improved stabilization of the CrCL-deficient stifle during the stance phase of gait with the EAI. However, even though peak relative tibial translation and peak relative tibial rotation were decreased, local maxima in tibial internal and external rotation, along with slight caudal translation, warrant additional investigation related to their influence on remaining stifle ligaments. Moreover, the reduction in peak relative tibial translation and rotation indicates improvement compared to RTTD/I at mid-stance.

To further investigate tibial translation and rotation, absolute tibial translation and rotation values were also assessed. Absolute tibial translation varied within ±1.2 mm for all stifle scenarios across the stance phase except at 50% stance in the CrCL-deficient and EAI-managed stifles. At 50% stance absolute tibial translation was 21.4 mm (cranial) and 2.1 mm (cranial) in the CrCL-deficient and EAI-managed stifles, respectively. Absolute tibial rotation in the CrCL-intact, CrCL-deficient, and EAI-managed stifles was primarily externally oriented. These findings are consistent with tibial kinematics reported in vivo for the CrCL-intact and CrCL-deficient stifles during the stance phase of gait. The EAI-managed stifle maintained consistent absolute external rotation (~4°) during early and late stance with decreased external rotation compared to the CrCL-intact stifle during mid-stance. Similar to the CrCL-intact stifle, peak external rotation occurs at 60% stance in the EAI-managed stifle. However, the EAI-managed stifle peak absolute external rotation was somewhat less than that of the CrCL-intact stifle. Nonetheless, stifle kinematics should be further investigated to evaluate EAI efficacy and to confirm effective stabilization of the CrCL-deficient stifle.

While this study found significant differences in model outcomes across the stance phase using comparative tests between the CrCL-intact, CrCL-deficient, and EAI-managed stifle, there is question whether these differences represent clinically significant differences. Future longitudinal investigation is needed to determine whether these differences are clinically relevant and whether they influence functional outcomes. The EAI was designed and developed to provide immediate and continuous stifle stabilization while eliminating stifle reconstruction necessary with osteotomy techniques. Therefore, this stabilization technique is less invasive than osteotomy techniques that alter bone morphology, and it does not require periarticular fibrosis for added stability necessary with extracapsular suture techniques. Extracapsular and osteotomy techniques were developed to provide optimal stifle stability during weight-bearing. Therefore, preoperative planning or surgical implementation is performed at stifle angles corresponding to mid-stance. However, extracapsular suture and osteotomy technique efficacy may vary across the gait cycle because of changes in suture tension and capability to neutralize tibial thrust, which are dependent on stifle angle. Biomechanical evaluation of the EAI in vivo or in vitro has not been reported to date. A case report of a mixed-breed Labrador retriever (5 years old, weighing 46 kg) that was non-weight bearing with a complete CrCL rupture, partial CaCL injury, and preoperative lameness score of 5/5 that was surgically managed using the EAI, described the dog was weight bearing, had a flexion–extension passive range of motion that ranged from 30° to 150°, and a lameness score of 0.5/5 at 12 weeks postoperative. Furthermore, abnormal cranial and caudal tibial translation and internal and external tibial rotation (qualitatively assessed) could not be elicited when the limb was evaluated through passive full range of motion. These preliminary findings suggest potential beneficial outcomes in 1 dog treated with the EAI but caution is required in interpreting these findings from a single dog. Longitudinal outcomes from a larger and diverse cohort of dogs over a longer period of time are needed to assess the EAI. In the present study, computer model-predicted findings indicated that the EAI-managed stifle decreased peak relative tibial translation and rotation during the stance phase of gait in the CrCL-deficient stifle. Although these computer model-predicted findings parallel preliminary case report findings, additional in vivo assessment of joint kinematics and surgical effectiveness during gait in dogs with EAI-stabilized stifle is needed.

In a previous study, using the same computer model of the canine pelvic limb, peak CaCL, lateral collateral ligament, and medial collateral ligament loads were found to increase by
88% BW, 640% BW, and 520% BW, respectively, in the TPLO-managed stifle compared to the CrCL-intact stifle (PL loads were not reported). Additionally, peak relative Tibial translation and relative tibial rotation after TPLO implementation (TPLO scenario relative to the CrCL-intact stifle) were 4.8 mm caudal and 5.2° internal rotation, respectively. Thus, based on an in silico comparative assessment for the same dog used to develop the computer model, EAI-managed stifle biomechanics differed from TPLO-managed stifle biomechanics during the stance phase of gait. However, a detailed comparative analysis across the entire stance phase of gait between the EAI-managed stifle and the TPLO-managed stifle was not conducted in this study.

Canine pelvic limb computer simulation models complement in vivo and in vitro studies to evaluate the biomechanical efficacy of CrCL deficiency stabilization procedures. However, as previously stated, to-date there are no in vivo or in vitro studies evaluating the EAI during gait. Therefore, canine pelvic limb computer model-predicted findings characterizing the EAI should be interpreted with consideration of the following limitations. The quasi-static computer model was used to evaluate discrete stance phase intervals and attempts to represent a complex biomechanical system based on a single dog. This model includes anatomically accurate bony morphology, experimentally determined ground reaction forces and moments, optimized muscle forces acting along linear force vectors, single-element ligamentous passive constraints, and stifle contact elements. Evaluation of intermediate discrete intervals (e.g., 5% stance, 15% stance, etc.) may lead to localized peak outcomes not reported in this study. Model-predicted outcomes may vary across breeds and individual dogs. Other dynamic tasks, such as trotting, running, turning, sit-to-stand, and jumping, were not evaluated, and the effects of these activities on EAI-managed stifle biomechanics were not determined. Clinically, planning for EAI component implantation is facilitated by the use of preoperative radiographs and identification of soft tissue landmarks. In this study, EAI components were appropriately sized for the morphology of the canine subject based on computed tomography imaging data. Preoperative planning and intraoperative execution may differ across individual patients based on morphological characteristics, surgeon experience, and measurement techniques leading to varying biomechanical outcomes. Meniscal elements in the computer model represented healthy, intact menisci, and the integrity of the menisci was maintained that did not reflect potential injury over time, which could result from CrCL deficiency. Additionally, the menisci were modeled as rigid bodies, which may not represent flexibility and deformation occurring in vivo. Meniscal release was not implemented in the model as it is not clinically performed. Moreover, this in silico canine pelvic limb model represented a fully healed stifle, free of functional deficits, surgical complications, and associated lameness. EAI components were assumed to maintain structural integrity and simulations did not assess the influence of fatigue loading on components; testing to assure no component failure occurs for the life cycle of the implant is necessary. Additionally, longitudinal in vivo changes in conditions, such as osteoarthritis, were not characterized in our canine pelvic limb model. Osteoarthritis commonly progresses in CrCL-deficient stifles regardless of surgical stabilization technique. Therefore, longitudinal study of joint health in the EAI-managed stifle is warranted.

In summary, comparative analysis of model-predicted outcomes indicated that biomechanics differed between the EAI-managed and CrCL-intact stifles. The analysis of model-predicted findings indicated that the EAI did not restore CrCL-intact stifle biomechanics, with CaCL and medial collateral ligament loading being significantly greater than that of the CrCL-intact stifle. However, patellar ligament loading in the EAI-managed stifle was not significantly different compared to the CrCL-intact stifle. When compared to the CrCL-deficient stifle, our comparative analysis indicated peak CaCL, lateral collateral ligament, and medial collateral ligament loading were significantly reduced and patellar ligament loading was not significantly different. Additionally, peak RTRD/EAI and RTTEAI were significantly reduced compared to RTRD and RTTG. Although investigation into long-term efficacy is necessary, this computer simulation study of the stance phase of gait suggests that the EAI surgical procedure appears to offer biomechanical advantage in a CrCL-deficient stifle.

ACKNOWLEDGMENTS

This study was supported by Embark Enterprises Inc. and New Generation Devices.

DISCLOSURE

Drs. Embleton and Barkowski are affiliated with Embark Enterprises Inc, which is the developer and patent holder of the extracapsular articulating implant system that stabilizes the stifle. The remaining authors declare no conflicts of interest related to this report.

REFERENCES


