Research in Veterinary Science 194 (2025) 105826

Contents lists available at ScienceDirect

Research in Veterinary Science

journal homepage: www.elsevier.com/locate/rvsc

Factors influencing stifle stability in canine limb press models: A systematic review and meta-analysis

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A R T I C L E I N F O

Keywords:
Canine limb press models
Periarticular forces
Canine stifle stability
CCL disease
Ex vivo models
Osteotomy techniques

A B S T R A C T

Limb press models have been widely used to explore aspects of canine stifle stability and to inform surgical practice. This systematic review aimed to synthesize test conditions, periarticular forces, and stability measures in canine limb press models for comparison with *in vivo* measurements. Six databases (EMBASE, Ovid MEDLINE, CAB abstracts, Agricola, Scopus, and Web of Science) were searched from inception to 17.03.2025. Peer-reviewed experimental studies using static, axially loaded, whole-limb constructs and reporting cranial tibial translation, rotation, and/or periarticular forces were eligible for inclusion. Twenty-three articles were included, reporting results from 368 healthy limbs of dogs of various breeds. Most (17/23) simulated stance during walking but landmarks for limb angulation were frequently poorly described or deviated from those used in cited studies. Cranial cruciate ligament transection caused mean cranial translation of 14.6 mm and mean internal rotation of 13.1°. Cranial translation and rotation were improved by osteotomy techniques, but data were heterogeneous, sample sizes were often small, and reporting standards inconsistent. Key data regarding quadriceps, gastrocnemius, or cranial thrust forces were extremely limited. This review identified potentially serious knowledge gaps regarding simulated muscle and periarticular forces, and concerns over limb angulation. Further evaluation of *ex vivo* models is warranted to determine the likely validity of currently reported models. If current models are shown to simulate muscle forces which do not reflect expected *in vivo* loading, this could help explain the difference between *in vivo* and *ex vivo* assessments of joint stability following surgeries such as TPLO and TTA.

1. Introduction

Cranial cruciate ligament (CrCL) disease is the most common non-traumatic condition of the canine stifle and surgical intervention is the most recommended treatment (Engdahl et al., 2021; Johnson et al., 1994).

Static limb press studies comprise most *ex vivo* biomechanical tests of canine stifle joint stability before and after CrCL transection and stabilization using various surgical techniques, since the first description by Warzee et al. (2001). These models simulate physiological periarticular loading during weight bearing by inducing secondary forces in cables and turnbuckles along the lines of action of key muscles. Canine limb press studies have largely reported good stability for tibial plateau levelling osteotomy (TPLO) (Kanno et al., 2019; Kim et al., 2009a, 2010; Kowaleski et al., 2005; Pozzi et al., 2006; Schmutterer et al., 2022; Warzee et al., 2001), tibial tuberosity advancement (TTA) (Apelt et al., 2007; Butler et al., 2011; Kim et al., 2009b; Schmutterer et al., 2023), and cranial tibial wedge ostectomy (CTWO) (Apelt et al., 2010). These data inform clinical decisions regarding target tibial plateau angle (TPA) for TPLO or required tuberosity advancement for TTA. However, *in vivo* data from radiographic (Kim et al., 2012; Skinner et al., 2013) and fluoroscopic studies (Rey et al., 2014; Schwede et al., 2018) demonstrating cranial tibial subluxation during weight bearing suggest that limb press results may be overoptimistic or misleading. In contrast, *ex vivo* models simulating muscle loads in fixed ratios yield stability rates more in line with *in vivo* observations, albeit with different osteotomy techniques (Jensen et al., 2020; Mazdarani et al., 2022).

Limitations of current models could include inappropriate limb angulations and positions, nonrepresentative loading, and poor simulation of periarticular muscle forces. Simulating only one or two muscle groups

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https://doi.org/10.1016/j.rvsc.2025.105826
Received 6 May 2025; Received in revised form 11 July 2025; Accepted 27 July 2025
Available online 5 August 2025

is insufficient to accurately represent *in vivo* loading. Ideally, a balanced simulation of agonistic and antagonistic muscle groups should be employed to test joint stability. The limb should be positioned to replicate the joint angulations associated with cranial tibial subluxation observed fluoroscopically. Additionally, the fixation of the femur to the testing rig may alter periarticular forces. Simulated muscle forces in a feline limb press inspired by canine models were highly dependent on femoral fixation method, such that ratios of quadriceps to gastrocnemius forces could deviate from reported *in vivo* values (Mazdarani et al., 2025).

This systematic review aims to identify reported bone and joint angles, vertical loads, simulated muscle loads, and periarticular forces in canine whole limb press models using static limb positioning and vertical loading. In addition, it aims to clarify experimental efficacy reported through magnitudes for cranial tibial thrust and/or numbers of CrCL-deficient limbs becoming stable following an intervention.

2. Methods

This systematic review followed the PRISMA-2020 guidelines (Page et al., 2021). Search strategy, analysis and inclusion criteria were prespecified in a written protocol (Supplementary material 1).

2.1. Information sources

Electronic databases including Ovid (EMBASE [1947-], Ovid MEDLINE [1946-], CAB abstracts and Agricola [1970-]), Scopus [1996-], and Web of Science [1900-] were searched on 28.02.2024. Results were exported to reference management software (Endnote 20, Clarivate, PA) for duplicate removal. An updated search was performed on 17.03.2025.

2.2. Search strategy

The search approach was refined through team discussions and with librarians from Copenhagen University Library. The search terms (dog OR canine) AND (cruciate) AND (cadaver OR "*ex vivo*" OR "*in vitro*") were utilised throughout all databases.

2.3. Inclusion and exclusion criteria

Peer-reviewed studies written in English, Portuguese, Danish, Norwegian, Swedish or German, with full text available were included. Conference abstracts, review articles, and book chapters were excluded to avoid the risk of double reporting or use of subjective data.

Additional criteria were: static limb press design; single limb position during each test; experimental canine cadaver studies; axial loading; use of simulated muscle forces; detailed methodological reporting. Use of different limb positions as separate sub-investigations within a study was acceptable. Constructs with dynamic testing, variable angle testing, partial limbs, other species or without axial loading were excluded from this study.

2.4. Screening process and selection of evidence

Screening was performed using Covidence systematic review software (Veritas Health Innovation, Melbourne, Australia, www.covidence.org). Title and abstract screening were performed by three independent reviewers (GVC, KHD, JEM) using the inclusion and exclusion criteria, with discrepancies resolved by a fourth reviewer (MBMN). Full-text screening was performed by two independent reviewers (GVC, JEM), with discrepancies resolved by discussion and consensus, involving a third reviewer (MBMN) as needed. Reasons for exclusion were noted.

2.5. Data extraction process and data items

Two independent reviewers (GVC, JEM) used a data extraction sheet within Covidence resolving disagreements through discussion or a third reviewer if needed.

Data extracted comprised, where available, joint angles (stifle and hock) and femoral angle, axial load, joint angle measurement method, stance phase simulated by model, femoral fixation, confirmation of angles during testing, simulated muscle forces, cranial translation, rotation, effect of CrCL transection and stabilizing techniques (if used) and measurements of cranial tibial thrust. Where standard deviations were not available, these were estimated from standard errors or ranges using *t*-values based on sample size. Subgroup data were combined by calculating group means and standard deviations where appropriate. Changes in tibial translation and internal rotation were calculated directly, with standard deviations imputed using the formula $\sqrt{\sigma_I^2 + \sigma_F^2 - 2(\rho \cdot \sigma_I \cdot \sigma_F)}$, where σ is standard deviation, I and F refer to the initial and final states, and the correlation coefficient ρ was empirically chosen as 0.8 (Higgins et al., 2024).

2.6. Synthesis of results

Data were tabulated, and selected data relating to tibial translation and rotation were used for meta-analysis of mean differences using R 4.4.2 (R Core Team, 2024) and packages *meta*, *metafor*, and *dmetar* (Balduzzi et al., 2019; Harrer et al., 2021; Viechtbauer, 2010). Criteria for meta-analysis were stifle joint angulation of 135°–145° (consistent with early to mid-stance when most cranial tibial subluxation occurs *in vivo*) and body mass ≥ 20 kg (to improve population homogeneity). Due to anticipated between-study heterogeneity, a random-effects model was used and the restricted maximum likelihood estimator was used to calculate the heterogeneity variance tau^2 in addition to the I^2 statistic (Viechtbauer, 2005). Knapp-Hartung adjustments were made to confidence intervals (Knapp and Hartung, 2003). Forest plots were used to visualize data, and outliers were screened and analyses repeated with selected studies excluded.

2.7. Bias assessment

Risk of bias was evaluated using a modified BOBQAT tool designed for cadaveric biomechanical laboratory research (Hohmann et al., 2024), evaluating each study on an 85-point scale. Items in the original version regarding bone density and cyclic loading were disregarded (Supplementary material 2). Assessments were performed by two independent reviewers (GVC, JEM) and discrepancies resolved by discussion and consensus.

3. Results

3.1. Study selection

Twenty-three studies were included in this review, and authors, titles and study objectives are reported in Supplementary material 3. Primary aims of the included studies could be broadly grouped as: stability assessments for TPLO (n = 4), TTA (n = 5), CWTO (n = 1), and lateral suture techniques (n = 2); femorotibial contact mechanics or pressure distributions (n = 6); interaction of muscle forces and CrCL deficiency or TPLO (n = 4); radiographic landmarks (n = 1). Database searches yielded 1055 records with three additional records identified by citation searching. After deduplication, 525 records remained: 467 were excluded during abstract and title screening, and 35 during full-text screening (Fig. 1, Supplementary material 3). One additional article was identified in the updated search (Sun et al., 2024), but excluded due to lack of numerical data. Partial or complete datasets were obtained from each included study.

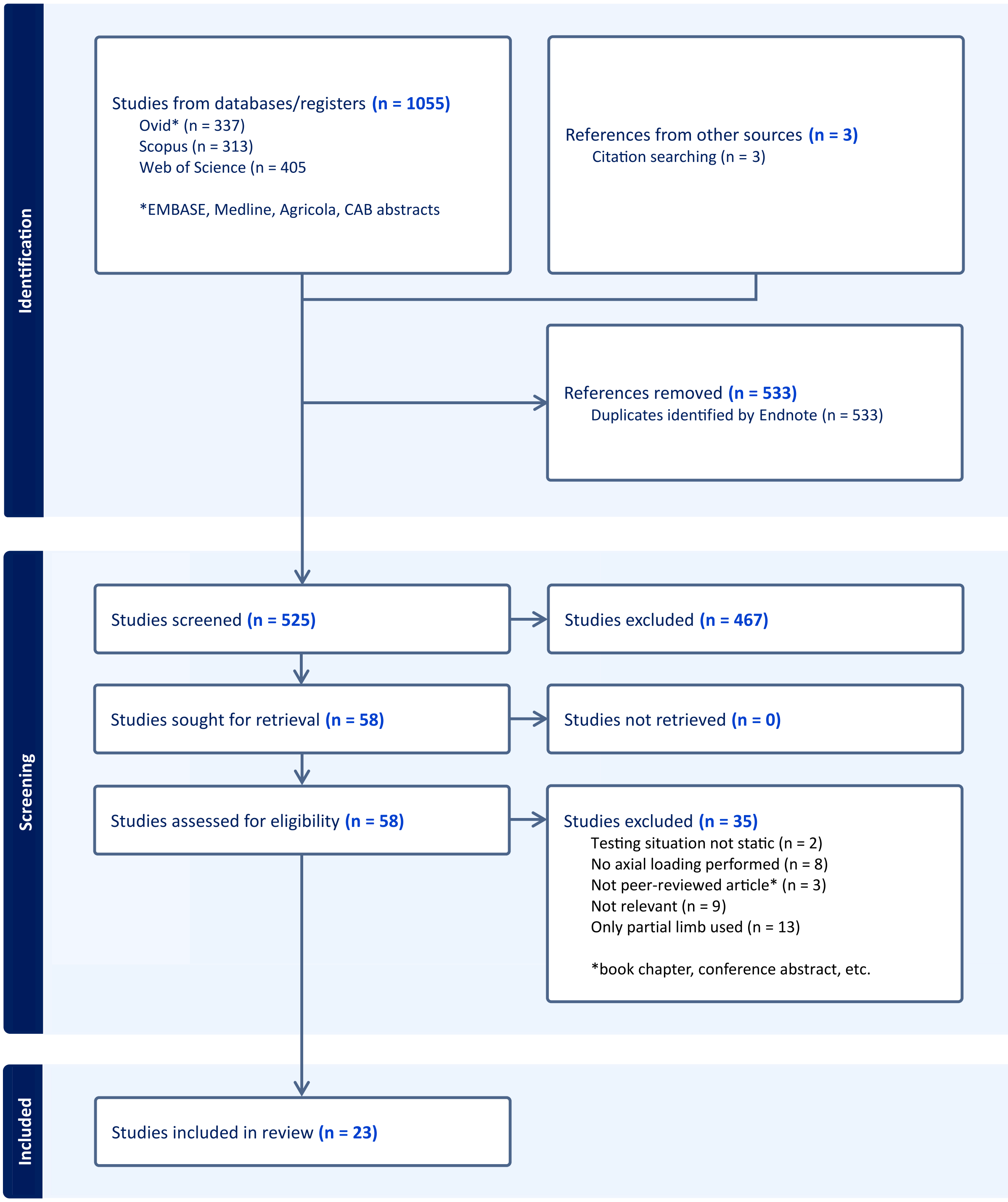

Fig. 1. PRISMA flowchart for this review.

3.2. Demographics

Demographic data including study location, number of dogs and limbs used, dog breeds, body mass and TPA are reported in Table 1. Ages were only reported by one study, but most studies specified skeletal maturity as an inclusion criterion.

No studies reported extreme TPA, defined as $>34^\circ$ (Duerr et al., 2007; Talaat et al., 2006), although two Beagle populations had high TPA values ($31.1 \pm 3.5^\circ$ and 30.3 ± 1.9) (Kanno et al., 2019; Kanno et al., 2012).

3.3. Limb press models

Limb press construction varied: custom-made limb presses were used in 13/23 studies, and materials testing machines in 10/23. Vertical loading was achieved using a weighted top-plate in 12/13 cases, and by a cable attached to the paw in the remaining study (Ober et al., 2022). Flexion-extension movement of the femur was permitted in one study (Pozzi et al., 2008) but prevented by the limb press in the remaining 22/23 studies. Nearly half (10/23) permitted unconstrained femoral rotation about its long axis (Aulakh et al., 2013; Butler et al., 2011; D'Amico et al., 2013; Kim et al., 2010; Kim et al., 2009b; Kim et al., 2009a; Pozzi et al., 2013; Pozzi et al., 2010; Schmutterer et al., 2023; Schmutterer

Table 1
Demographics of the included studies. Body mass and initial tibial plateau angles are reported as either mean (standard deviation), range, mean (range), or mean (standard deviation; range).

Study	Country	Study design	Dogs (n)	Limbs (n)	Breed	Body mass (kg)	TPA (°)
Warzee et al. (2001)	United States	Non-randomized	NR	15	NR	27–36	26.5 (1.9)
Kowaleski et al. (2005)	United States	Randomized	6	12	NR	20–25	NR
Pozzi et al. (2006)	United States	Non-randomized	31	62	Various breeds	25–35	24–30
Apelt et al. (2007)	United States	Non-randomized	NR	10	Various breeds	26.7 (24.1–30)	26.2 (23–29)
Pozzi et al. (2008)	United States	Non-randomized	12	24	Various breeds	25–35	NR
Kim et al. (2009a)	United States	Non-randomized	8	8	NR	32 (3)	23 (1)
Kim et al. (2009b)	United States	Non-randomized	8	8	NR	32 (3)	23 (1)
Pozzi et al. (2010)	United States	Non-randomized	8	8	NR	28–35	NR
Kim et al. (2010)	United States	Non-randomized	8	16	NR	32 (3)	23 (1)
Apelt et al. (2010)	United States	Non-randomized	3	6	NR	25–30	22.2 (1.3; 20.5–24.0)
Hoffmann et al. (2011)	United States	Randomized	15	30	Mixed breed	27.4 (4.3)	25.8 (3.9; 17–30)
Butler et al. (2011)	United States	Non-randomized	9	18	NR	22.8 (4.6)	23.0 (1.9)
Plesman et al. (2012)	Canada	Non-randomized	23	23	Medium to large breeds	22.83 (5.34)	NR
Kanno et al. (2012)	Japan	Non-randomized	8	8	Beagle	9.8 (1.1)	31.1 (3.5)
Pozzi et al. (2013)	United States	Non-randomized	4	8	NR	32 (3)	23 (1)
D'Amico et al. (2013)	United States	Non-randomized	11	20	Various breeds	23.4 (4.6)	NR
Aulakh et al. (2013)	United States	Randomized	16	32	Mixed breed	26.8 (4.6)	NR
Ramirez et al. (2015)	Belgium	Non-randomized	7	14	NR	21–37	23.43 (2.2)
Drew et al. (2018)	Australia	Non-randomized	6	6	Greyhound	30.3 (2.37)	25 (0.89)
Kanno et al. (2019)	Japan	Non-randomized	20	20	Beagle	9.98 (1)[a]	30.0 (2.1)[c]
Ober et al. (2022)	Israel	Non-randomized	6	6	NR	20–25	24 (1)
Schmutterer et al. (2022)	Germany	Non-randomized	7	7	Retrievers	31.5 (4.1)[b]	21.3 (1.9)
Schmutterer et al. (2023)	Germany	Non-randomized	7	7	Retrievers	25–40	NR

NR – not reported.
[a] combined mean and standard deviation from two populations: 10.5 (0.9) and 9.4 (0.9).
[b] median reported instead of mean, with standard deviation.
[c] combined mean and standard deviation from two populations.

et al., 2022).

3.4. Gait, stance phase and applied load

Walking gait informed limb positioning and loading in most (17/23) studies (Table 2), with gait information unreported in 3/23 studies.

Early stance phase was variably represented with stifle and hock joint angles of 137°–145° and 135°–143°, respectively. Mid-stance was consistently represented by stifle and hock angles of 135° and 145°, respectively. Sources included kinematic studies (Fischer et al., 2018;

Table 2
Target angles, verification methods, tested gait and vertical loads for canine limb press models. Target angles for the femur (or coxofemoral joint), stifle joint and hock joint are reported as point targets or as ranges, except when clearly specified in the text as being mean (standard deviation). Method of determining bone and joint angles is specified along with necessary landmarks (if given). Loads are expressed as percentage body weight unless otherwise stated.

Study	Femur (°)	Stifle (°)	Hock (°)	Method	Landmarks	Stance phase	Gait	Load
Warzee et al. (2001)	70	135	145	T	NR	Mid	Walk	30 % (33 %)[a]
Kowaleski et al. (2005)	70	145	135	R	NR	Early	Walk	30 %
Pozzi et al. (2006)	70	135, 100–110	NR	G	FD, TD	NR	NR	20 %
Apelt et al. (2007)	70	130–140	140–150	R	NR	Mid	Walk	30 %
Pozzi et al. (2008)	NR	145	NR	NR	NR	NR	NR	200 N
Kim et al. (2009a)	NR	130–140	130–140	R	FD, TD	Mid	Walk	30 %
Kim et al. (2009b)	NR[b]	135	NR[b]	R	NR	NR[b]	Walk	30 %
Pozzi et al. (2010)	NR	130–140	130–140	G	NR	Mid	Walk	30 %
Kim et al. (2010)	NR[b]	85–95	85–95	G	NR[b]	NR	Deep flexion	30 %
Apelt et al. (2010)	70	145	135	R	NR	Early	Walk	30 %
Hoffmann et al. (2011)	NR[c]	140–150, 130–140	130–140, 140–150	G, R	NR[c]	Early, Mid	Walk	30 %
Butler et al. (2011)	113[d], 120[d], 135[d]	145, 135, 125	140, 145, 160	G	FD, TD, MD	Early, Mid, Late	Walk	30 %
Plesman et al. (2012)	70	135	NR	G	FD, TD	NR	NR	20 %
Kanno et al. (2012)	115 (5)[d]	137 (5)	129 (5)	R	NR	At PVF	Walk	65.3 %
Pozzi et al. (2013)	NR	135, 90	135, 90	G	FD, TD	Mid, Flexed	Walk and flexed	30 %
D'Amico et al. (2013)	103[d], 115[d], 133[d]	145, 135, 125	140, 145, 155	G	FD, TD	Early, Mid, Late	Walk	30 %
Aulakh et al. (2013)	148[e], 160[e], 178[e]	145, 135, 125	140, 145, 155	G	FD, TD, MD	Early, Mid, Late	Walk	33 %
Ramirez et al. (2015)	70	132–142	138–148	G	KL	Early	Walk	50 %
Drew et al. (2018)	NR	135	145	G	NR	Mid	Walk	30 %
Kanno et al. (2019)	115 (5)[d]	137 (5)	129 (5)	R	NR	At PVF	Walk	65.3 %
Ober et al. (2022)	NR	135, 125, 105	NR	G	NR	NR	Walk	0.5–5.0 kg
Schmutterer et al. (2022)	NR	135	140	U	NR	At PVF	Trot	30 %
Schmutterer et al. (2023)	NR	135, 145	145, 135	U	NR	Mid, Late	Trot	30 %

NR – not reported in text, T – angular templates, R – radiography, G – goniometer, U – ultrasound technique, FD - femoral diaphysis, TD - tibial diaphysis, MD - metacarpal diaphysis, KL – kinematic landmarks, PVF – peak vertical force.
[a] Discrepancy between text and figure.
[b] Refers to Kim et al., 2009a).
[c] Refers to Apelt et al. (2007).
[d] Coxofemoral joint angle (pelvic angle 32° (5°) for both Kanno et al. (2012) and Kanno et al. (2019)).
[e] Angle relative to vertical axis.

Hottinger et al., 1996; Korvick et al., 1994), authors' own studies (Kanno et al., 2012), and other studies in this review. Two studies did not specify a source (Pozzi et al., 2013; Pozzi et al., 2008). Femoral angulation was 70° in 7/23 studies (2/7 mid-stance, 3/7 early stance, 2/7 not reported). Five studies reported coxofemoral angulation, but only two used an intact hemipelvis (Kanno et al., 2019; Kanno et al., 2012), and reference landmarks for hip joint angulation were unclear for the remaining three (Aulakh et al., 2013; Butler et al., 2011; D'Amico et al., 2013). Joint angles during loading were not always specified, and showed increases, decreases, and no change (Table 3).

Landmarks for confirming angulation were not reported in 15/23 studies, although 2/15 referred to prior reports from the same group. One study used kinematic landmarks (Ramirez et al., 2015), while remaining studies referenced the diaphyses of two or three limb segments.

Fifteen studies cited sources for the applied vertical load: 8/15 cited Warzee et al. (2001) which referred to a kinetic and kinematic study of walking foxhounds reporting peak vertical forces of 20–40 % body weight on the hind limbs (Page et al., 1993). Other sources included own studies (Kanno et al., 2019, Kanno et al., 2012), biomechanical studies (Holden et al., 1994; Shahar and Banks-Sills, 2004), and an earlier study by the same group (Kim et al., 2009a).

3.5. Muscle forces

Methods of simulating muscle forces varied (Table 4). Spring characteristics were not reported for the 5/22 studies combining a spring and turnbuckle. Quantitative data were rarely reported and as either raw values or normalized to vertical load, complicating direct comparisons due to differing units, vertical loads, and dog sizes and conformations.

Table 3
Femoral and joint angles under load. Reported angles for the femur (coxofemoral joint), stifle joint and hock joint under each study's vertical load conditions, as mean (standard deviation) and range, along with joint situation (if specified). Corresponding target angles are given in Table 4. Values for Kanno et al. (2019) are calculated means (standard deviation) for two combined groups.

Study	Femur (°)	Stifle (°)	Hock (°)
Apelt et al. (2007)	NR	136.1 (2.0)	144.5 (2.5)
Pozzi et al. (2008)	NR	*ca.* 105[b]	NR
Kim et al. (2009a)	NR	141 (2)[b], 141 (5)[c], 142 (4)[d]	NR
Kim et al. (2009b)	NR	141 (3)[b], 142 (4)[c], 143 (3)[d]	NR
Kim et al. (2010)	NR	105.5 (6.2)[b], 107.5 (4.0)[c], 107.6 (6.0)[d]	NR
Apelt et al. (2010)	NR	144.3 (2.5); 140.2–149.6	134.2 (2.7); 130.0–139.7
Kanno et al. (2012)[a]	114.7 (3.8)[b], 114.7 (2.9)[c]	137.6 (4.1)[b], 132.9 (2.2)[c]	130.8 (4.7)[b], 121.5 (1.0)[c]
Pozzi et al. (2013)	NR	138 (3.5)[b], 142.7 (2.8)[c], NR (flexed)	NR
Ramirez et al. (2015)	70 (1.0)[b], 70 (1.0)[c], 70 (1.0)[d]	120.6 (4.1)[b], 105.9 (1.9)[c], 127.1 (16.4)[d]	112.9 (5.8)[b], 107.4 (3.0)[c], 129.6 (18.1)[d]
Drew et al. (2018)	NR	136 (2.8)	143 (1.5)
Kanno et al. (2019)[a]	121.2 (5.0)[a]	136.2 (1.3)[b], 137.4 (1.6)[d]	128.2 (1.8)[b], 127.8 (1.5)[d]
Schmutterer et al. (2023)	NR	134.9 (0.6), 145.3 (0.4)	NR

NR – not reported.
[a] Coxofemoral joint angle (pelvic angle 31.9° (5.6°) and 30.2° (2.7°) for Kanno et al. (2012) and Kanno et al. (2019), respectively).
[b] Intact joint situation.
[c] Cranial cruciate ligament transected.
[d] Stabilizing surgery (*e.g.* TPLO, TTA).

Conversion to body weight normalized values, including linearly scaling to compensate for different vertical loads, yielded quadriceps values of 0.99–1.39 for intact limbs at 30 % body weight loads (Supplementary material 4). Values following cranial cruciate ligament transection and TPLO were inconsistent.

3.6. Cranial tibial thrust force

Only one study reported cranial tibial thrust forces under load, using a turnbuckle and spring linkage between the caudal tibia and a load cell (Hoffmann et al., 2011). Reported data were limited to a single limb in graphical form and indicated maximum thrust force of approximately 7.5 N.

3.7. Translational displacements

Most studies (17/23) reported cranial tibial positioning relative to the femur, with varying methods and measures of absolute or relative displacement (Table 5). Apart from radiography, methodologies included electromagnetic tracking systems that detect position and orientation of sensors relative to a source transmitter (Aulakh et al., 2013; Butler et al., 2011; D'Amico et al., 2013), construction of 3D models from CT images for measuring rotation using software (Kim et al., 2010; Kim et al., 2009a; Kim et al., 2009b; Pozzi et al., 2013), radiographic markers (Warzee et al., 2001), and an ultrasound system (Schmutterer et al., 2023; Schmutterer et al., 2022).

CrCL transection increased cranial tibial translation, with a pooled estimate of 14.6 mm (95 % CI: 11.8 mm; 17.3 mm) across 13/23 studies included in the meta-analysis (Fig. 2A). Target stifle joint angles were 135° or 137° in 10/13 and 145° in 3/13 studies. Data from 4/17 studies were excluded due to low body mass, low stifle joint angle, or use of estimated marginal means (D'Amico et al., 2013; Kanno et al., 2019; Kanno et al., 2012; Kim et al., 2010). Heterogeneity was high ($tau^2 =$ 17.6 (95 % CI: 8.8; 60.2) and $I^2 = 94.3$ % (95 % CI: 91.9 %; 96.0 %), and three studies were identified as possible outliers (Apelt et al., 2010; Pozzi et al., 2006; Warzee et al., 2001).

Stabilization with either TPLO or TTA reduced or eliminated cranial tibial translation. TPLO to 5°–7° reduced cranial tibial translation to −0.1 mm (95 % CI: −3.4 mm; 3.2 mm) in 5/13 studies (Fig. 2B). Heterogeneity was again high ($tau^2 = 6.7$ (95 % CI: 2.2; 59.2) and $I^2 = 98.7$ % (95 % CI: 98.0 %; 99.1 %), and three studies were identified as outliers (Apelt et al., 2010; Pozzi et al., 2006; Warzee et al., 2001). TTA to a patellar tendon angle of 90° reduced cranial tibial translation to 1.4 mm (95 % CI: −0.9 mm; 3.6 mm) in 3/13 studies (Fig. 2C). Heterogeneity was lower for this group with $tau^2 = 0.33$ (95 % CI: 0.0; 29.5) and $I^2 =$ 31.7 (95 % CI: 0.0 %; 92.9 %), but one potential outlier was identified graphically (Apelt et al., 2007).

Removal of potential outliers did not change these estimates appreciably (Supplementary material 5).

3.8. Rotational displacements

Rotational movements at the stifle joint were reported in 12/23 studies, with 12/23 reporting internal/external rotation, 7/23 reporting adduction/abduction or varus/valgus, and 5/23 reporting flexion/extension data. Movements were determined as above, or with projection of marker pins (Warzee et al., 2001), and measures of absolute or relative internal or external rotation of the tibia relative to the femur varied between studies (Table 6). CrCL transection was associated with internal rotation of the tibia of 13.1° (95 % CI: 8.8°; 17.3°) using 9/12 studies (Fig. 3 A). Data from 3/12 studies were excluded due to low body mass or low stifle joint angle (Kanno et al., 2019, Kanno et al., 2012; Kim et al., 2010). Outlier removal did not change this pooled estimate appreciably (Supplementary material 5).

Stabilization procedures reduced internal rotation. TPLO data from 2/12 studies indicated a mean internal rotation of 5.1° (95 % CI: −50.1°;

Table 4
Methods of simulating hind limb muscles and associated forces. Muscle forces are reported as mean (standard deviation) in N for Kim et al. (2009a) and Drew et al. (2018), with standard deviations calculated from the 95 % confidence intervals for the mean for the latter, and as values normalized to the applied load of 65.3 % body weight for Kanno et al. (2012, 2019).

Study	Muscles	Linkage	Quadriceps	Gastrocnemius	Hamstrings
Warzee et al. (2001)	Q, G	TS, T			
Kowaleski et al. (2005)	Q, G	TS, T			
Pozzi et al. (2006)	Q	T			
Apelt et al. (2007)	Q, G	TS, T			
Pozzi et al. (2008)	Q	T			
Kim et al. (2009a)	Q, G	T, T	436.7 (192)[a], 653.8 (168)[b], 496.1 (202.2)[c]		
Kim et al. (2009b)	Q, G	T, T			
Pozzi et al. (2010)	Q, G	T, T			
Kim et al. (2010)	Q, G	T, T			
Apelt et al. (2010)	Q, G	TS, T			
Hoffmann et al. (2011)	Q, G	TS, T			
Butler et al. (2011)	Q, G	T, T			
Plesman et al. (2012)	Q	T			
Kanno et al. (2012)	Q, G, H	T, T, T	3.3 (0.5)[a], 4.0 (0.6)[b]	2.2 (0.3)[a], 2.3 (0.4)[b]	0.4 (0.0)[a], 0.8 (0.1)[b]
Pozzi et al. (2013)	Q, G	T, T			
D'Amico et al. (2013)	Q, G	T, T			
Aulakh et al. (2013)	Q, G	T, T			
Ramirez et al. (2015)	Q, G	T, T			
Drew et al. (2018)	Q, G	T, T	308 (45)[a], 313 (56)[c], 393 (47)[d]		
Kanno et al. (2019)	Q, G, H	T, T, T	3.5 (0.3)[b], 3.6 (0.5)[c]	1.9 (0.3)[b], 2.1 (0.3)[c]	0.4 (0.0)[b], 0.4 (0.0)[c]
Ober et al. (2022)	Q	W			
Schmutterer et al. (2022)	Q, G	T, T			
Schmutterer et al. (2023)	Q, G	T, T			

Q – quadriceps, G – gastrocnemius, H – hamstrings, T – turnbuckle and cable, S – spring, W – weighted cable.
[a] Intact stifle.
[b] Cranial cruciate ligament transection.
[c] Tibial plateau levelling osteotomy.
[d] Over-rotated tibial plateau levelling osteotomy.

60.4°), with high heterogeneity (tau^2 = 35.3, I^2 = 93.3 % (95 % CI 78.0 %; 98.0 %) (Fig. 3B) reflecting the disparity between these two studies. TTA data from 3/12 studies were more uniform with a point estimate of 1.4° (95 % CI: −0.9°; 3.6°), tau^2 = 0.3 (95 % CI: 0.0; 29.5), I^2 = 31.7 % (95 % CI: 0.0 %; 92.9 %) (Fig. 3C).

3.9. Bias assessments

Risk of bias assessments for the included studies are presented in Table 7. Only one study reported a sample size calculation, for detection of cranial tibial subluxation (Plesman et al., 2012). Statistical analyses frequently lacked evaluation of test assumptions. In 5/11 studies using both hind limbs (Apelt et al., 2010; Kowaleski et al., 2005; Pozzi et al., 2013; Pozzi et al., 2008; Ramirez et al., 2015), paired limbs were treated statistically without apparent adjustment for a repeated effect of dog: 6/11 studies used a mixed model statistical approach that could adjust for paired limbs, although this was not explicitly stated (Aulakh et al., 2013; Butler et al., 2011; D'Amico et al., 2013; Hoffmann et al., 2011; Kim et al., 2010; Pozzi et al., 2006).

Sources of funding were clearly reported in only 15/23 studies.

3.10. Confidence in the level of evidence

Confidence in the evidence presented here is affected by the lack of direct clinical data and the inherent limitations of laboratory-based investigations, which are generally considered to comprise low levels of evidence. Despite consistent directions of change observed in the meta-analysis, inconsistency (study heterogeneity), limited numbers of studies and limbs in the sub-groups, and indirectness (healthy joints, *ex vivo* studies), result in low confidence in the magnitude of these changes relative to the *in vivo* situation (Schünemann et al., 2013).

4. Discussion

This systematic review evaluated 23 studies regarding static canine limb press models. However, few studies reported data that may be considered fundamental to an assessment of their validity and clinical relevance. The choice of joint or bone angles, axial loads and muscle simulation appeared inconsistent in some cases. Descriptions often lacked detail, and justifications were frequently weak. Thorough methodological reporting is crucial for maintaining scientific integrity, enabling scrutiny, and ensuring transparency and clarity. Preventing flexion-extension movement of the femur and allowing unconstrained femoral rotation about its long axis are crucial methodological aspects which should be consistently reported and justified in future studies.

Ex vivo studies have inherent limitations and weaknesses. Removal of soft tissues and supporting structures, natural *postmortem* changes, effect of freezing and thawing, differing study populations, and the lack of a standardized protocol for testing or reporting, make comparisons between different *ex vivo* models and between *ex vivo* and *in vivo* models challenging. Simplifying numerous muscles, ligaments and tendons to two simulated muscle groups in most models reduces validity, given the important role of CrCL agonist muscles in determining dynamic joint stability (Adrian et al., 2013). Additionally, while using healthy joints unaffected by CrCL disease and associated fibrosis improves standardization, these joints do not reflect the chronic degenerative process most often seen clinically (Rafla et al., 2025). Femoral and joint angulations varied slightly within simulated stance phases, but the impact of this on either periarticular forces or validity of results is unclear. A significant issue is the discrepancy between landmarks for determining joint angles and those used in cited kinematic studies. For example, widely cited landmarks of the greater trochanter, between the lateral epicondyle and fibular head, lateral malleolus, and distolateral aspect of the fifth metatarsal bone to determine stifle and hock angles (Hottinger et al., 1996) do not match those based on diaphyseal axes. On average, angles defined using femoral and tibial diaphyses were 21° lower (Jensen et al., 2020). Landmarks employed were often unreported, and only one study specifically used kinematically derived axes. Some studies probably used more extended stifle joints than intended, shifting the line of action of the axial load relative to the stifle joint centre of rotation and altering

Table 5
Reported cranial tibial translations under study conditions. Values are given as mean (standard deviation) in mm, unless otherwise stated. Where separate groups of limbs were tested within a study, this is noted under group (G).

Study	G	Intact	CrCLx	MMR	Procedure	Comment
Warzee et al. (2001)		0	18.9 (3.4)	–	−6.3 (1.8)	TPA 0°
					−3.2 (0.8)	TPA for stability under nominal load
					−8.9 (1.1)	TPA 0° and CaCLx
Kowaleski et al. (2005)	1	0	16.33 (2.18)	–	14.78 (1.96)	Distally centred TPLO to target TPA 5°
	2		14.93 (4.68)		2.98 (2.76)	Correctly centred TPLO to target TPA 5°
Pozzi et al. (2006)	1a	0	–	1.9 (1.02)	2.04 (1)	Intact CrCL, MMR then MCPHM
	1b		28.2 (9.1)	6.08 (1.43)	6.13 (1.63)	CrCLx, MMR then MCPHM
	2a	0	NR	1.7 (0.9)	2.4 (1.1)	TPLO to target 5°–7°, MMR and MCPHM
	2b			6.6 (1.9)	7.4 (2.5)	TPLO with unchanged TPA, MMR and MCPHM
Apelt et al. (2007)		0	11.1 (3.4)	–	−2.9 (1.3)	TTA of 16.2 mm (3.2 mm)
					2.3 (2.2)	TTA of 10.2 mm (3.7 mm)
Kim et al. (2009a)		6.6 (2.1)	21.9 (3.1)	–	4.6 (2.3)	TPLO to target TPA 6°
Kim et al. (2009b)		9.0 (1.6)	24.5 (2.2)	–	9.7 (3.0)	TTA to PTA of 90°
Kim et al. (2010)	1	8.0 (1.8)	10.8 (5.5)	–	1.4 (2.5)	TPLO to target TPA 6°
	2	10.2 (1.3)	12.7 (3.1)	–	10.1 (2.5)	TTA to PTA of 90°
Apelt et al. (2010)		0	8.5 (1.5)	–	5.7 (1.7)	CWTO with wedge of TPA-5°
					4.8 (2.1)	CWTO with wedge of TPA°
					1.4 (3.3)	CWTO with wedge of TPA + 5°
					−1.1 (3.1)	CWTO with wedge of TPA + 7.5°
Butler et al. (2011)		−3.5 (2.8)	10.3 (7.1)	−2.8 (5.6)	−2.0 (3.3)	SJA 125°, MMR after TTA to PTA of 90°
		−2.4 (3.9)	9.0 (3.1)	0.07 (4.9)	−1.5 (6.7)	SJA 135°, MMR after TTA to PTA of 90°
		0.02 (2.0)	7.6 (4.6)	−0.5 (4.2)	0.2 (5.0)	SJA 145°, MMR after TTA to PTA of 90°
Kanno et al. (2012)		0	2.1 (1.3)	–	7.2 (2.3)	CrCLx under hamstring load:
					7.5 (2.3)	procedure is load released and then reapplied
Aulakh et al. (2013)	1	0	17.1 (5.5)	–	3.0 (5.7)	SJA 125°, LS to cranial extensor groove
			14.1 (2.5)		7.0 (4.1)	SJA 135°, LS to cranial extensor groove
			12.2 (2.6)		5.1 (4.2)	SJA 145°, LS to cranial extensor groove
	2	0	15.8 (3.6)	–	−0.2 (5.6)	SJA 125°, LS to tibial tuberosity
			14.5 (4.6)		3.3 (5.1)	SJA 135°, LS to tibial tuberosity
			13.3 (2.0)		3.6 (4.4)	SJA 145°, LS to tibial tuberosity
Pozzi et al. (2013)		10.2 (3.9)	25.1 (4.9)	–	–	SJA target 135°
		10.4 (3.6)	11.0 (4.0)			SJA target 90°
D'Amico et al. (2013)		−3.84 (0.25)[a]	−2.52 (0.26)[a]	–	−3.53 (0.26)[a]	SJA 125°, KYON LS system
		−1.80 (0.26)[a]	0.17 (0.26)[a]		−1.28 (0.26)[a]	SJA 135°, KYON LS system
		−0.40 (0.26)[a]	1.38 (0.26)[a]		0.29 (0.26)[a]	SJA 145°, KYON LS system
Ramirez et al. (2015)		0.5 (0.5)	11.4 (3.1)	–	5.5 (5.4)	Procedure is quadriceps pre-activation
Kanno et al. (2019)	1	0	12.0 (1.4)	–	−1.9 (1.7)	Ideally centred TPLO to target TPA 6.5°
	2		11.5 (2.9)		−5.3 (1.1)	Distally centred TPLO to target TPA 6.5°
Schmutterer et al. (2022)		−1.3 (0.5)[b]	12.2 (1.1)[b]	–	−1.4 (0.6)[b]	TPLO to target TPA 6°
					−1.5 (0.9)[b]	TPLO to target TPA 1°
Schmutterer et al. (2023)		−1.7 (0.6)	12.6 (4.5)	–	−0.5 (2.0)	SJA 135°, TTA 12 mm
					−1.2 (0.4)	SJA 145°, TTA 12 mm
					4.6 (6.9)	SJA 135°, TTA 9 mm
					0.3 (1.3)	SJA 145°, TTA 9 mm

CaCLx – transected caudal cruciate ligament; CrCL – cranial cruciate ligament; CrCLx – transected CRCL; CWTO – closing wedge tibial ostectomy; Intact – intact cranial cruciate ligament; LS – lateral suture; MCPHM – medial caudal pole hemimeniscectomy; MMR – medial meniscal release; procedure – additional surgical intervention (see comments); NR – not reported; PTA – patellar tendon (ligament) angle; SJA – stifle joint angle; TPA – tibial plateau angle; – - not specifically tested.
[a] Estimated marginal (least squares) means reported.
[b] Standard deviations calculated from 95 % confidence intervals for the mean using an appropriate two-tailed value for t at $P = 0.05$.

induced muscle forces. This shift can inadvertently increase or decrease simulated muscle forces due to changes in joint torques as axial loading shifts cranially or caudally relative to the joint centre of rotation. Cranial tibial thrust force will also increase or decrease in concert with stifle joint angle changes, impacting the validity of reported results.

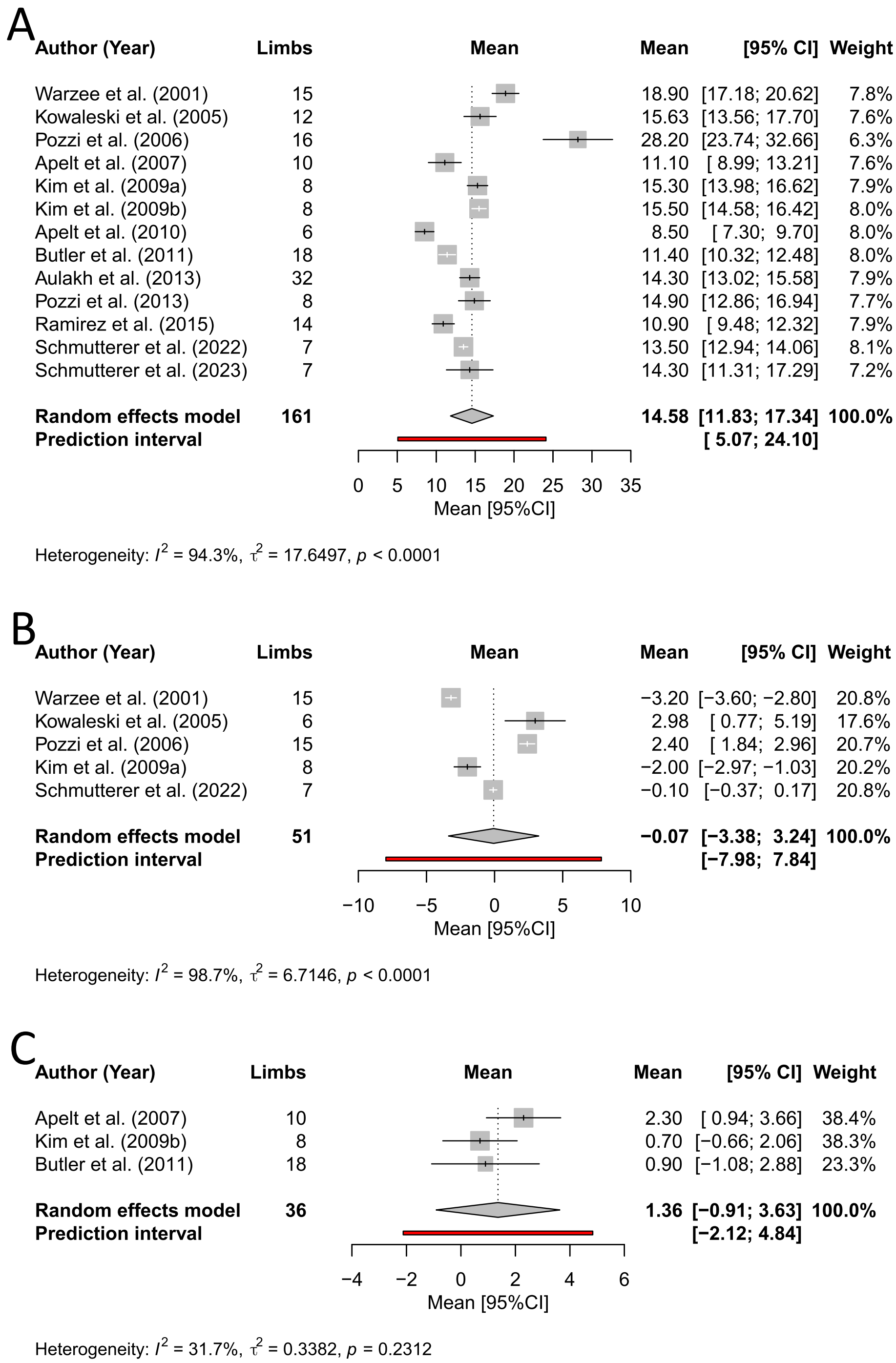

Fig. 2. Reported cranial tibial translations. Data are presented as change from intact status (reported or imputed data) following A - cranial cruciate ligament transection, B - tibial plateau levelling osteotomy, and C - tibial tuberosity advancement. Individual results for all studies reporting relevant data are shown with number of limbs, mean translation and study weighting. Positive values indicate cranial translation, negative values caudal translation. An overall estimate using a random effects model (diamond) and prediction interval (red bar) are provided underneath, along with estimates of data heterogeneity. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

The disparity in reported quadriceps forces for similar sized dogs could be due to differences in conformation, lever arm length or testing conditions. The relationship between applied vertical load, quadriceps force and femoral fixation method in canine models remains unknown, but feline studies suggest a linear relationship between 10 % – 40 % body weight and a negative effect of restricting flexion-extension (Mazdarani et al., 2025). We are not aware of canine *in vivo* assessments of either quadriceps or patellar ligament force. Feline studies have suggested peak patellar ligament forces of 80 N and 50–60 N during walking, corresponding to *ca.* 1.8 N/N when normalized to body weight (Hasler et al., 1997). Caprine studies have reported mean peak patellar ligament forces of 801 N with a mean peak vertical force of 249 N during walking, corresponding to *ca.* 1.3 N/N when normalized (Korvick et al., 1996). Apart from one study (Kim et al., 2009a), calculated values were generally below these. However, patellar ligament forces and quadriceps forces are not necessarily equivalent, as a lever arm effect of the patella has been reported in human studies (van Eijden et al., 1986). Current *in silico* canine hind limb models are limited by assumptions,

Table 6
Reported tibial internal rotations under study conditions. Values are given as mean (standard deviation) in degrees, unless otherwise stated. Negative values indicate external rotation relative to the femur. Where separate groups of limbs were tested within a study, this is noted under group (G).

Study	G	Intact	CrCLx	MMR	Procedure	Comment
Warzee et al. (2001)		0	23.2 (6.8)	–	9.5 (6.3)	TPA 0°
Kim et al. (2009a)		−5.4 (5.3)	8.9 (6.3)	–	−4.6 (7.3)	TPLO to target TPA 6°
Kim et al. (2009b)		−5.0 (2.9)	9.3 (4.4)	–	−1.8 (5.3)	TTA to PTA of 90°
Kim et al. (2010)	1	4.7 (4.5)	1.4 (8.5)	–	−2.8 (5.7)	TPLO to target TPA 6°
	2	1.8 (4.1)	0.1 (5.8)	–	−2.9 (8.5)	TTA to PTA of 90°
Butler et al. (2011)		−1.1 (1.5)	0.9 (3.0)	0.8 (3.2)	0.14 (3.4)	SJA 125°, MMR after TTA to PTA of 90°
		−1.5 (2.0)	2.4 (3.9)	1.1 (3.0)	0.5 (3.3)	SJA 135°, MMR after TTA to PTA of 90°
		−0.7 (1.4)	0.5 (3.6)	0.14 (2.2)	0.3 (3.1)	SJA 145°, MMR after TTA to PTA of 90°
Kanno et al. (2012)		0	4.7 (1.7)	–	18.9 (5.1)	CrCLx under hamstring load: procedure is load released and then reapplied
					18.1 (4.9)	
Aulakh et al. (2013)	1	0	12.57 (4.86)	–	−5.95 (2.51)	SJA 125°, LS to cranial extensor groove
			11.4 (4.31)		−4.48 (3.58)	SJA 135°, LS to cranial extensor groove
			11.2 (4.82)		−0.89 (4.18)	SJA 145°, LS to cranial extensor groove
	2	0	14.02 (3.43)	–	−6.85 (1.80)	SJA 125°, LS to tibial tuberosity
			12.31 (3.52)		−8.16 (3.44)	SJA 135°, LS to tibial tuberosity
			11.53 (3.31)		−5.31 (3.35)	SJA 145°, LS to tibial tuberosity
Pozzi et al. (2013)		−13.8 (3.6)	4.6 (5.7)		–	SJA target 135°
		−10.7 (3.6)	8.3 (4.2)			SJA target 90°
D'Amico et al. (2013)		−13.90 (2.58)[a]	−2.21 (2.58)[a]	–	−14.86 (2.58)[a]	SJA 125°, KYON LS system
		−13.87 (2.58)[a]	−2.22 (2.58)[a]		−14.99 (2.58)[a]	SJA 135°, KYON LS system
		−12.64 (2.58)[a]	−2.67 (2.58)[a]		−13.61 (2.58)[a]	SJA 145°, KYON LS system
Kanno et al. (2019)	1	0	17.5 (2.7)	–	6.2 (3.0)	Ideally centred TPLO to target TPA 6.5°
	2		16.3 (4.9)		2.2 (5.7)	Distally centred TPLO to target TPA 6.5°
Schmutterer et al. (2022)		−0.4 (0.4)[b]	7.9 (2.5)[b]	–	0.4 (0.2)[b]	TPLO to target TPA 6°
					0.5 (0.3)[b]	TPLO to target TPA 1°
Schmutterer et al. (2023)		−0.7 (0.8)	12.1 (7.2)	–	−0.2 (1.1)	SJA 135°, TTA 12 mm
					−0.3 (1.1)	SJA 145°, TTA 12 mm
					3.7 (6.4)	SJA 135°, TTA 9 mm
					−1.0 (0.8)	SJA 145°, TTA 9 mm

CrCLx – transected cranial cruciate ligament; CWTO – closing wedge tibial ostectomy; Intact – intact cranial cruciate ligament; LS – lateral suture; MCPHM – medial caudal pole hemimeniscectomy; MMR – medial meniscal release; procedure – additional surgical intervention (see comments); NR – not reported; PTA – patellar tendon (ligament) angle; SJA – stifle joint angle; TPA – tibial plateau angle; – - not specifically tested.
[a] Estimated marginal (least squares) means reported.
[b] Standard deviations calculated from 95 % confidence intervals for the mean using an appropriate two-tailed value at $P = 0.05$ for t (2.365).

and data for quadriceps and gastrocnemius muscle forces appear inconsistent with *in vivo* data from other quadrupeds (Brown et al., 2020; Shahar and Banks-Sills, 2004). Further studies, both *in vivo* and *ex vivo*, are required to establish construct validity concerning simulated muscle forces in canine models.

The limited data on cranial tibial thrust (Hoffmann et al., 2011) appears inconsistent with peak *in vivo* measurements of CrCL loads during walking which range from 63 to 124 N in goats weighing 503–636 N (Holden et al., 1994), assuming a linear relationship between body weight and CrCL load. This may represent methodological issues such as use of a spring in the measurement coupling, or a failure of limb press models to generate realistic forces. Accurate determination of cranial tibial thrust in these models is essential for validation and informing both future research and clinical practice, highlighting a significant knowledge gap.

Observed cranial tibial translations following CrCL transection were large but consistent with our clinical observations of acute traumatic CrCL rupture. While the two studies using smaller dogs were excluded from the meta-analysis (Kanno et al., 2019; Kanno et al., 2012), observed translations were only slightly smaller than the estimated mean and within the prediction interval, indicating greater relative laxity in smaller dogs. Values reported *in vivo* for 18 clinical cases of CrCL rupture with body mass 20–40 kg and mild to moderate osteoarthritis were 9.7 mm (SD 2.7 mm) (Tinga et al., 2018), within our calculated prediction interval but lower than the estimated mean. This is consistent with periarticular fibrosis providing partial stabilization of the stifle joint. Following TPLO *ex vivo*, both mild caudal and cranial translations were reported, with our calculated prediction interval consistent with an *in vivo* study demonstrating caudal translation of 3.4–4.9 mm in 10/16 joints and cranial translation of 3.8–4.4 mm in 5/16 joints (Tinga et al., 2020). However, both these *in vivo* and *ex vivo* reports are inconsistent with reports of gross instability during walking of 3/5 TPLO-operated joints (Rey et al., 2014). Following TTA, slight cranial translation was identified *ex vivo*. Fluoroscopic studies have identified gross instability during walking in 4/6 joints and 9/10 joints operated with TTA (Rey et al., 2014; Schwede et al., 2018). Quantitative *in vivo* measurements were not available, but *ex vivo* results appear inconsistent with these observations.

Observed internal rotation following CrCL transection was generally large, and again the studies using smaller dogs (Kanno et al., 2019, Kanno et al., 2012) were consistent with the values for larger breeds obtained in the meta-analysis. Values reported *in vivo* indicate increased internal rotation for CrCL rupture cases of 7.0° – 11.0° during stance

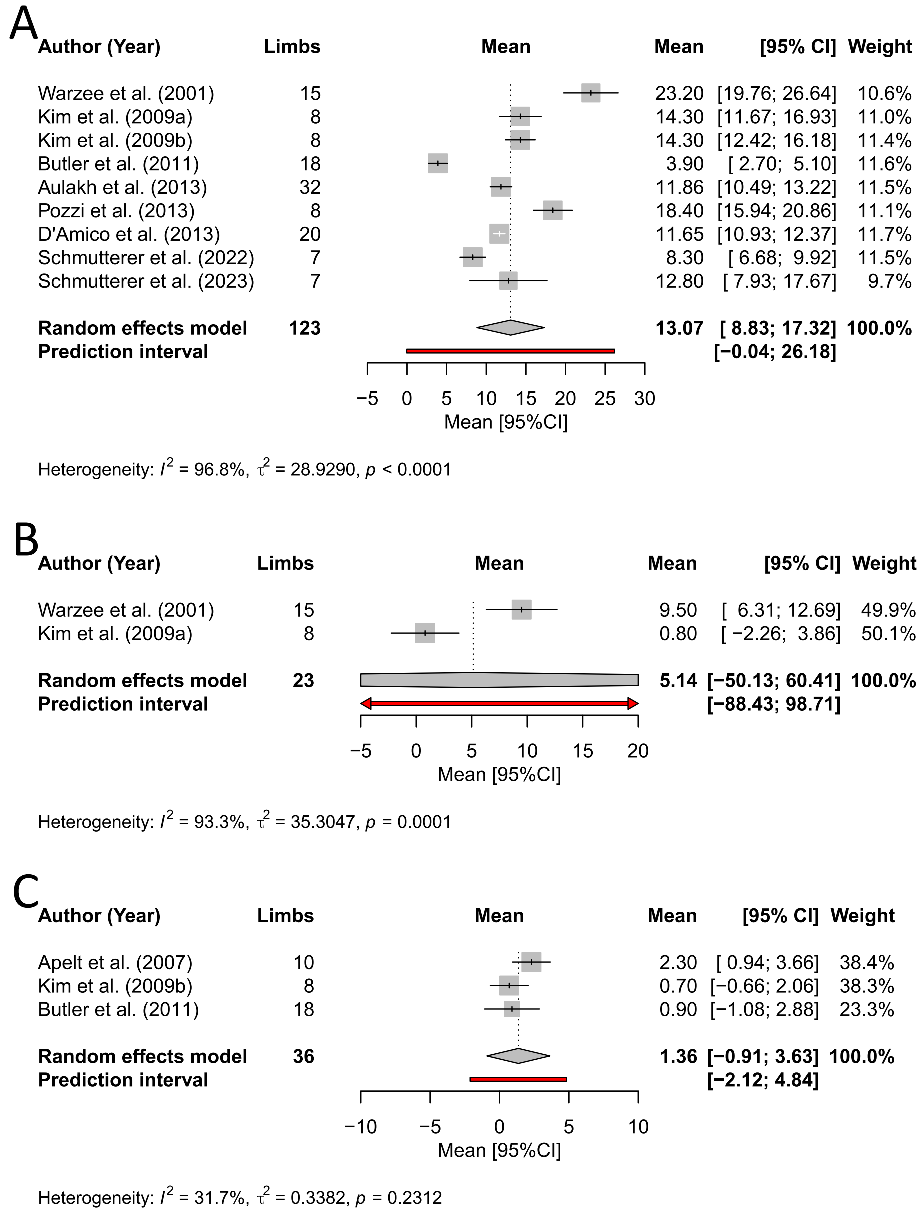

Fig. 3. Reported cranial tibial rotations. Data are presented as change from intact status (reported or imputed data) following A - cranial cruciate ligament transection B - tibial plateau levelling osteotomy, and C - tibial tuberosity advancement. Individual results for all studies reporting relevant data are shown with number of limbs, mean translation and study weighting. Positive values indicate internal rotation, negative values external rotation. An overall estimate using a random effects model (diamond) and prediction interval (red bar) are provided underneath, along with estimates of data heterogeneity. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

phase compared to intact joints, with internal rotation reducing by 7.0° – 11.8° following TPLO (Tinga et al., 2020). The two *ex vivo* studies reported widely disparate results, resulting in an extreme prediction interval, which hinders comparison. We are not aware of *in vivo* data for joint rotation after TTA. Our results for both translation and rotation may have been negatively influenced by marked heterogeneity. Larger studies and more consistent data are needed to increase confidence in these estimates, especially regarding the effect of osteotomy techniques.

Common potential sources of bias were limited reporting of demographics, lack of randomization, statistical analyses, and reporting of financial support. The latter may be related to journal guidelines and thus unfairly harsh. Accurate demographic reporting would facilitate comparison and assessments of whether results can be extended to other populations.

Data were inconsistently reported and in varying formats. Our meta-analyses are limited by the available data and in particular the need to calculate change from baseline where this was not stated. An empirical choice for the correlation coefficient between paired comparisons of 0.8 was selected, however insufficient data to calculate or directly estimate this value were found. Higher and lower values could reduce or increase the imputed standard deviation for change from baseline and thus the estimated mean translations or rotations and associated prediction intervals. We assumed a strong correlation between measured translation in the intact and CrCL transected joints, related to landmark separation and joint size. Similarly, we expected that joints with relatively higher degrees of internal rotation when intact would also exhibit higher internal rotation following transection due to general joint laxity. The meta-analyses were in broad agreement with limited clinical data, allowing for the non-diseased state of the joints used in these studies, but some uncertainty remains. Studies varied in size, with wide variations in number of limbs used and only 3/23 could not be considered small (<30 limbs), increasing the risk of outlier bias in these data. Relatively few

Table 7
Bias assessments for the studies included in this review. Assessments were performed using a modification of the Biomechanics Objective Basic Science Quality Assessment Tool (BOBQAT) score (see supplementary file), with two questions excluded (bone density, cyclic testing) resulting in a maximum total score of 85. Each study's individual scores, total score, and percentage score is shown. Studies are identified by first author and year of publication.

Study	Q1	Q2	Q3	Q4	Q5	Q6	Q7	Q8	Q9	Q10	Q11	Q12	Q13	Total	%
Warzee et al. (2001)	10	5	5	5	10	7	7	0	3	5	5	5	5	72	85
Kowaleski et al. (2005)	10	3	3	5	10	7	7	0	3	0	3	5	0	56	66
Pozzi et al. (2006)	10	3	3	5	10	7	7	0	3	5	5	5	5	68	80
Apelt et al. (2007)	10	3	3	5	10	7	7	0	3	5	3	5	5	66	78
Pozzi et al. (2008)	10	3	3	5	10	4	7	0	5	5	5	5	5	67	79
Kim et al. (2009a)	10	3	5	5	10	7	7	0	3	5	5	5	0	65	76
Kim et al. (2009b)	10	3	3	3	10	4	4	0	3	5	5	5	0	55	65
Apelt et al. (2010)	10	3	3	5	10	7	7	0	3	5	5	5	5	68	80
Kim et al. (2010)	10	3	3	3	10	4	7	0	3	5	5	5	0	58	68
Pozzi et al. (2010)	10	3	3	3	10	7	7	0	3	3	5	5	0	59	69
Butler et al. (2011)	10	3	3	5	10	7	7	0	3	5	5	5	0	63	74
Hoffmann et al. (2011)	10	5	3	3	10	7	7	0	3	3	5	5	5	66	78
Kanno et al. (2012)	10	5	3	5	10	10	10	0	3	5	5	5	0	71	84
Plesman et al. (2012)	10	3	3	5	10	4	7	5	5	5	5	5	5	72	85
Aulakh et al. (2013)	10	3	3	5	10	7	7	0	3	5	5	5	5	68	80
D'Amico et al. (2013)	10	3	3	5	10	7	7	0	3	3	3	5	5	64	75
Pozzi et al. (2013)	10	3	3	5	10	7	7	0	3	5	5	5	5	68	80
Ramirez et al. (2015)	10	3	3	5	10	10	10	0	5	5	5	5	5	76	89
Drew et al. (2018)	10	5	3	3	10	7	7	0	5	5	3	5	5	68	80
Kanno et al. (2019)	10	5	3	5	10	10	10	0	5	5	5	5	0	73	86
Ober et al. (2022)	10	3	5	5	10	4	4	0	0	5	5	5	5	61	72
Schmutterer et al. (2022)	10	5	5	5	10	7	7	0	3	5	5	5	5	72	85
Schmutterer et al. (2023)	10	5	5	5	10	7	7	0	3	5	5	5	5	72	85
Maximum score	10	5	5	5	10	10	10	5	5	5	5	5	5	85	100

Q1 - Clear and answerable purpose statement; Q2 - Demographic information about specimens is included; Q3 - Condition of specimens described/checked; Q4 - Description of surgical technique, if appropriate; Q5 - Appropriate outcome measures selected; Q6 - Loading conditions selected to limit inherent bias; Q7 - Magnitude of loads applied consistent with clinical conditions; Q8 - Sample size calculation performed; Q9 - Statistical analysis appropriate; Q10 - Results consistent with methods and testing protocol; Q11 - Confounders/limitations mentioned and discussed; Q12 - Conclusions based on data/ results only; Q13 - Industry funding and potential conflicts of interest declared.

studies were available for meta-analysis of either TPLO or TTA. No studies reported numbers of limbs achieving stability for comparison with fluoroscopic data. Observer bias in some reported measures cannot be excluded, in addition to known issues with observer errors in measurement.

Database selection was based on human and veterinary recommendations (Bramer et al., 2017; Grindlay et al., 2012), but we cannot be certain that all relevant studies were included. Narrower search terms resulted in detection of few of the final included studies, indicating issues with identifying limb press models *via* title, abstract and keyword. Typically, systematic reviews utilize randomized studies, and only 3/23 included studies met this criterion. However, the data of interest to this review was not predicated on randomization, and the authors believe that the use of all included studies was valid.

Future research should consider use of a standardized reporting format, *e.g.* based on the modified BOBQAT tool used here, to ensure all relevant data are reported. These should include full demographic information for assessment of applicability to other populations, detailed descriptions of limb press design or construction, the prevention or allowance of femoral flexion-extension movement and long axis rotation, as well as periarticular forces generated within the model. Use of randomized studies should be encouraged, along with open access to data *via* repositories.

5. Conclusion

This review has identified a potentially serious knowledge gap regarding the forces achieved during axial loading of canine limb constructs, as well as concerns over limb angulation. The lack of standardized protocols, inconsistent methodologies, and questionable validity of simulated muscle forces are critical flaws which hinder clinical utilization of published results. Evaluation of *ex vivo* models which examine both the effect of joint angulation and of construct fixation method, along with the response to varying axial load, are warranted to determine the likely validity of currently reported models. If these models are shown to simulate muscle forces which do not reflect expected *in vivo* loading, this could help explain the difference between *in vivo* and *ex vivo* assessments of joint stability following surgeries such as TPLO and TTA.

Authors contribution

Glauco Vinício Chaves, James Edward Miles and Leonardo Augusto Lopes Muzzi conceived the review topic. Database searches were performed by James Edward Miles. Glauco Vinício Chaves, James Edward Miles, Kathrine Højte Dahl and Michelle Brønniche Møller Nielsen performed screening and discrepancy resolution. Glauco Vinício Chaves and James Edward Miles extracted relevant data, performed analysis, and wrote the first draft of the manuscript. All authors commented on previous versions and read and approved the final manuscript.

Ethics approval

This is a systematic review of current literature, and no ethical approval is required.

Funding

Institutional funding was received for this study and there was no influence on study design, collection or analysis or interpretation of data, writing, or decision to submit.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests:

James Edward Miles reports a relationship with Nordic Veterinary Orthopaedic Society that includes: board membership. If there are other authors, they declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.rvsc.2025.105826.

Data availability

Extracted data and data used for the meta-analysis are available at: https://doi.org/10.6084/m9.figshare.28749980

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