The Effect of Stifle Angle on Stifle Kinematics following TPLO: An *in vitro* Experimental Analysis

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Objective: To determine the ability of the Tibial Plateau Leveling Osteotomy (TPLO) to restore normal joint kinematics in a cranial cruciate ligament (CrCL)-deficient stifle through a loaded range of motion.

Methods: Paired pelvic limbs from 12 dogs were compared in an *in vitro* biomechanical study. Each limb was placed in a custom designed jig at 120° of stifle extension under an axial load of 20% body weight. Electromagnetic motion tracking sensors were placed on the distal femur and proximal tibia. A force was applied at approximately 10 N/sec to mimic the action of the quadriceps muscle. Force application allowed the limb to move from 120° to maximal extension. Positional data was acquired at 60 points/second. Each limb was tested under normal, CrCL-deficient, and TPLO-treated conditions.

Results: The TPLO failed to normalize CTT within the CrCL-deficient stifle; however, values trended towards intact values throughout the range of motion. No significant differences were noted in internal rotation in any of the three conditions from 120° – 137°. Hyperextension values did not differ significantly between conditions.

Conclusion: Data from this biomechanical model suggests that the TPLO fails to neutralize CTT throughout a loaded range of motion. Internal rotation and hyperextension values were not found to differ significantly between intact, CrCL-deficient and TPLO repaired stifles. The effectiveness of the TPLO in restoring normal biomechanics is more significant at greater angles of flexion.

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INTRODUCTION

The cranial cruciate ligament (CrCL) is known to prevent cranial tibial translation (CTT), limit excessive internal rotation, and prevent hyperextension of the stifle [1]. Deficiency of this ligament leads to three-dimensional kinematic changes in the stifle joint throughout the entire gait cycle [2]. With an intact CrCL, motion during the swing phase occurs in three planes while in the stance phase motion is limited to flexion and extension. Rupture of this ligament alters stifle motion such that motion in three planes is present during both the swing and stance phases. Additionally there is significant CTT that occurs throughout the stance phase [2,3]. These changes in gait kinematics are thought to alter the load distribution across the articular surfaces of the femur and tibia promoting the development of osteoarthritis [4,5]. The ideal surgical technique to address a CrCL-deficient stifle should restore all three functions to the stifle throughout the entire gait cycle and prevent the progression of osteoarthritis. The complex and dynamic functions of the CrCL has made the development of a consistently effective surgical technique challenging which is evident by the vast array of proposed surgical techniques as well as the failure of a single technique to prove superior in the reestablishment of normal limb function [6,7] or prevention of the progression of osteoarthritis [8].

The tibial plateau leveling osteotomy (TPLO) is a commonly performed technique for repair of a CrCL-deficient stifle. The TPLO is based on the theory that the caudal slope of the tibial plateau promotes cranial translation of the tibia during weight bearing. The TPLO neutralizes this slope resulting in pure compression without CTT during loading [9]. The ability of the TPLO to eliminate this CTT at a standing angle has been demonstrated in three separate biomechanical cadaveric studies [10-12]. While these studies validate the principle function of the TPLO, only one study accurately addresses the effect of the TPLO on limiting excessive internal rotation [12]. In that study, CTT and internal rotation were evaluated in intact, CrCL-deficient and TPLO repaired stifles. While marked CTT and internal rotation were noted in the CrCL-deficient stifle, no significant differences were noted between intact stifles and TPLO repaired stifles, suggesting the TPLO restores two of the three functions of the CrCL at a standing angle.

While these biomechanical TPLO studies validate the normalization of the relationship of the femur and tibia at a standing angle, the vital assessment of the three-dimensional changes throughout weight bearing has not been performed. Two *in vivo* experimental studies have described the three-dimensional kinematic changes that occur in the absence of the CrCL [2,3]. While both studies documented the presence of CTT, changes in rotation were conflicting. Additionally a single *in vitro* biomechanical study that evaluated the three-dimensional changes throughout a range of motion in an intact and CrCL-deficient stifle found that from 30° to 95° of flexion there were no significant differences in CTT or internal rotation between the two conditions [13]. However, the limbs were not loaded to mimic weight-bearing, making comparisons to other studies at loaded angles difficult. While each of these studies evaluates a range of motion, none assess the effect of the TPLO.

To the authors' knowledge no study has concurrently evaluated the effect of the TPLO on CTT, internal rotation and hyperextension through a loaded range of motion.

CHAPTER I: Literature Review

A: Canine Stifle Anatomy

The canine stifle joint is a complex, synovial joint that allows motion in three planes. While the primary movement of the stifle joint is flexion-extension the anatomy of the joint also allows for cranial-caudal translation, internal-external rotation, varus-valgus angulation, medial-lateral translation and compression-distraction [2]. It is composed of 3 long bones: distal femur, proximal tibia, proximal fibula; 4 sesamoid bones: popliteal, medial fabella, lateral fabella, patella; 2 menisci; 4 major ligaments: medial collateral, lateral collateral, cranial cruciate (CrCL), caudal cruciate; a joint capsule; and multiple muscles [14]. Each of these components works in concert to contribute to the dynamic stability of the stifle joint.

The distal femur has two major articulations formed from three articular surfaces. The femoropatellar articulation is formed by the cranially positioned trochlear groove and the patellar sesamoid. This articulation dynamically lengthens the lever arm of quadriceps muscle during motion improving the efficiency of extensor function [15]. The femorotibial articulation is formed by distinct lateral and medial condylar articulations with the tibia. This is the major weight bearing articulation. Axially the condyles each articulate directly with the tibial plateau. Abaxially the condyles articulate with the respective menisci. While both condyles are convex in the sagittal and transverse planes the convexity of the lateral condyle is more pronounced [16]. This asymmetry assists in the relatively asymmetric motions of each condyle during flexion and extension; the lateral condyle is able to move more caudally with respect to the articular surface of the tibia as the stifle progresses through flexion.

The proximal tibia, also termed the tibial plateau, is composed of medial and lateral condyles, which are separated by a medial and lateral intercondylar eminence. Each condyle is concave transversely, convex sagitally and sloped in a caudal direction when viewed from the side [16]. The caudolateral aspect of the tibial surface has a small facet for articulation with the head of the fibula. The fibula contributes little to weight bearing and primarily functions as a site for muscular and ligamentous attachments [14].

The medial collateral ligament originates on the medial epicondyle of the femur, fans out as it extends distally forming strong attachments with the joint capsule and medical meniscus and inserts on the proximal medial aspect of the tibia. This ligament is taut in both flexion and extension. The lateral collateral ligament originates on the lateral epicondyle of the femur, extends distally forming a loose attachment to the joint capsule, and inserts on the head of the fibula. This ligament is taut in extension but progressively loses tension with stifle flexion. The primary function of the collateral ligaments is to limit varus and valgus motion of the tibia [17]. This role is most pronounced in stifle extension. Additionally, the collateral ligaments are the primary restraint against rotation in extension [18].

The CrCL originates on the medial aspect of the lateral femoral condyle and extends craniodistally to insert on the cranial intercondylar region of the tibia. It is composed of a craniomedial and caudolateral band. During extension both bands are taut; in flexion only the craniomedial band is taut. The CrCL prevents cranial displacement of the tibia with respect to the femur, limits internal rotation in conjunction with the caudal cruciate ligament and prevents hyperextension [1]. The caudal cruciate ligament originates on the lateral aspect of the medial femoral condyle and extends caudodistally to insert on the medial aspect of the popliteal notch of the tibia. It is composed of cranial and caudal bands that are much less distinct than the bands of the CrCL. The cranial portion is taut in flexion but lax in extension; the caudal portion is taut in extension and loose in flexion [1]. The caudal cruciate ligament prevents caudal translation of the tibia with respect to the femur and limits internal rotation through a twisting sequence with the CrCL.

The major blood supply to the cruciate ligaments originates from branches of the genicular artery [17]. These branches within the synovial tissues that ensheath the cruciate ligaments penetrate the ligament transversely to supply the ligaments. The infrapatellar fat pad and tissues caudal to the joint capsule also contribute. The lack of direct vascularization leaves the mid section of the ligaments without a substantial blood supply.

Mechanoreceptors and nerve fibers are also present within the cruciate ligaments. These have an important role in triggering reflex arcs to protect the knee from excessive strains. Activation of these receptors within the CrCL is known to result in simultaneous contraction of the caudal thigh muscles and relaxation of the cranial thigh muscles [14] thus helping to coordinate the muscular moments contributing to stifle stability and coordinated motion. These mechanoreceptors are unevenly distributed throughout the CrCL with the highest numbers being present in the proximal third and the lowest numbers in the distal third [19].

The menisci are C-shaped fibrocartilaginous structures positioned between the femoral and tibial condyles. They are composed of a body and two horns, cranially and caudally. The horns are wedge shaped in cross section. Peripherally they are thick and convex but become thinner and concave as they slope centrally. Each meniscus has a cranial and caudal meniscotibial ligament that firmly adheres it to the tibia. Additionally, the intermeniscal ligament attaches the two menisci cranially. The lateral meniscus also has an attachment to the femur, the meniscofemoral ligament. The meniscal horns are vascular and richly innervated in contrast to the body, which lacks significant blood vessels and nerves. Only the peripheral 10 – 15% of the body receives a quantifiable blood supply from the synovial plexus extending from the joint capsule [17]. The remainder of the body receives nutrients from the synovial fluid. The menisci function in load transmission, deepen the articular surface of the tibial plateau to provide joint stability, improve joint congruity, aid in joint lubrication and the rich supply of nerve endings within the horns provide proprioceptive information to help coordinate muscular adaptations to joint load [17]. During flexion of the stifle the caudal horns of the menisci move caudally. Due to the more intimate attachment of the medial meniscus to the joint capsule and the lack of a meniscofemoral ligament the medial meniscus is less mobile than the lateral meniscus. Thus during flexion the lateral meniscus allows for relatively more caudal transposition of the lateral femoral condyle [17].

The combined anatomical features of the articular surface and ligamentous components of the stifle results in what has been termed the screw home mechanism [17]. In the extended stifle the medial and lateral collateral ligaments are taut. The craniomedial and caudolateral bands of the CrCL are similarly taut. The medial and lateral condyles of the femur are seated on the articular surface of the tibia cushioned peripherally by the menisci. As the stifle flexes the lateral collateral ligament becomes lax while the medial collateral ligament remains taut. This allows for the lateral condyle to move caudally with respect to the tibial plateau creating relative internal rotation of the tibia. As the condyle moves caudally the lateral meniscus is also displaced caudally to prevent impingement. The caudolateral band of the CrCL relaxes and the two cruciate ligaments twist upon themselves to limit the degree of internal rotation. As the stifle is again extended the lateral condyle moves cranially as the lateral collateral ligament tightens thus restoring the initial ligamentous and positional states. This dynamic arrangement allows the tibia to rotate internally during the swing phase of gait.

In addition to the bony and ligamentous components of the stifle many muscles contribute to stifle motion. These muscles can be broadly categorized into extensors and flexors. However, many of the muscles can assist in either stifle flexion or extension depending on the anatomical region of the muscle that is activated. The primary extensor of the stifle is the quadriceps femoris. This muscle is composed of 4 separate muscles: the rectus femoris, vastus lateralis, vastus intermedius and vastus medialis. While the rectus femoris arises from the ilium the remaining three muscles originate from the proximal femur. All four muscles converge to insert on the tibial tuberosity via the patellar ligament. Contraction of this muscle group forces the patella into the trochlear groove and pulls the tibial crest cranially, thus extending the stifle.

The biceps femoris muscle also contributes to stifle extension. It is composed of a cranial and caudal head. The cranial head has a broad origin from the sacrotuberous ligament and ischiatic tuberosity. It runs distally and cranially converging at the patella to insert on the tibial tuberosity via the patellar ligament. This orientation allows the cranial head to function in both stifle flexion and extension. Unlike the cranial head, the caudal head only assists in stifle flexion. This head originates at the ischiatic tuberosity. It runs distally and cranially along the lateral surface of the limb to directly insert on the lateral aspect of the tibial tuberosity. A portion of this insertion also continues distally to insert on the calcanean tuberosity. This anatomical arrangement contributes to stifle flexion and hock extension.

Like the biceps femoris the sartorius muscle has distinct components that function in opposition with respect to motion at the stifle joint. The cranial portion of the sartorius originates from the iliac crest and inserts with the quadriceps muscles on the patella. This portion of the muscle contributes to stifle extension. Conversely, the caudal portion contributes to stifle flexion. It originates on cranioventral ilium and inserts on the cranial border of the medial tibia where it assists in stifle flexion.

The semitendinosus muscle functions in stifle flexion and hock extension. This muscle originates from the ischiatic tuberosity and runs distally and medially to insert on the proximal medial tibia. A portion of the insertion also continues distally to contribute to the common calcanean tendon. Contraction of this muscle leads to stifle flexion, hock extension and hip extension.

Another important stifle flexor is the semimembranosus. This muscle is composed of a cranial and caudal head. Both heads originate at the ischiatic tuberosity. The cranial head runs distally to insert on the distal femur. This portion of the muscle does not contribute to stifle flexion; it main role is in hip extension. The caudal head runs distally and medially to insert deep to the medial collateral ligament. Activation of this muscle promotes stifle flexion and hip extension.

The gracilis plays an important role in limb adduction. However, it also has fibers that contribute to stifle flexion as well as hip and hock extension. This muscle originates at the pelvic symphysis. It extends distally to insert on the cranial border of the tibia and, in conjunction with the semitendinosus, on the calcanean tuberosity.

The gastrocnemius does not play a direct role in flexion or extension of the stifle [14]. However, it's anatomical positioning effects stifle stability. This muscle originates from the medial and lateral supracondylar tuberosities of the femur and inserts on the calcanean tuberosity. Contraction of this muscle leads to extension of the hock. In addition, contraction creates a caudodistally directed force on the distal femur. In an intact stifle the CrCL counteracts this force. In the CrCL-deficient stifle this force contributes to the cranial translation of the tibia with respect to the femur.

Finally, the popliteus muscle plays a unique role in the stifle joint. It originates from the lateral condyle of the femur and inserts on the proximocaudal surface of the tibia. While this muscle functions in stifle extension, unlike the other muscles it also promotes internal rotation of the tibia with respect to the femur [14].

B. Pathophysiology of Cranial Cruciate Disease

Rupture of the cranial cruciate ligament (CrCL) is one of the most commonly recognized orthopedic conditions in dogs [20]. In a small number of dogs rupture of the CrCL occurs acutely after a traumatic event. This type of rupture often occurs in dogs less than 4 years of age [17] and is commonly seen in combination with other ligamentous injuries to the caudal cruciate and medial collateral ligaments. Isolated traumatic injury to the CrCL has been reported in puppies. In puppies a traumatic event leads to avulsion of the ligament, most commonly at the tibial attachment site [21]. While truly acute ruptures do occur, they are relatively rare. The chronic disease process far exceeds the number of acute ruptures. The clinical history at presentation may include an acute onset of lameness but physical examation findings and radiographic degenerative changes often support the chronic nature of the disease process.

Deterioration of the cruciate ligament has been shown to be a natural process that occurs with age [22]. In the undisrupted state the CrCL is composed of type I, and to a lesser extent, type III, collagen fibrils. These fibrils are divided into smaller subfibrils and microfibrils [23] and have a crimped arrangement. This crimping allows for elongation of the fibrils when exposed to a tensile load. The major cell type in the cruciate ligament is the fibroblast. Three different types of fibroblasts have been identified: fusiform, ovoid and spheroid [24]. With age this normal anatomy undergoes significant change: the overall population of fibroblasts within the ligament decrease, some fibroblasts undergo metaplasia to form chondrocytes and the functional arrangement of the fibrils becomes disrupted [22]. This degenerative process is typically more pronounced in the central core of the ligament, a region that is known to lack a significant vascular supply. As ligament degeneration is a natural biologic process in dogs, failure of the ligament is likely attributable to other factors as not every aging dog develops cranial cruciate disease. However, the exact cause or causes still remains elusive. Many different contributing factors have been proposed.

A breed predisposition has been reported in certain breeds including the Rottweiler, Labrador retriever, Chesapeake Bay retriever, Newfoundland, Akita, Neopolitan mastiff, Saint Bernard and Staffordshire bull terrier [25,26]. These breeds may have conformational characteristics that differ from dogs less prone to rupture. In a study comparing the mechanical properties of CrCLs in rottweilers and greyhounds it was shown that rottweilers required half the load per unit body weight for rupture to occur as compared to the greyhound [27]. Anatomical variation in the proximal aspect of the tibia has also been proposed as an influential factor in CrCL rupture. Some authors contend that excessive caudal sloping of the articular surface of the tibia, caused either by anatomy or posture, exposes the ligament to higher strains during loading contributing to chronic degeneration and eventual rupture [28-31]. However, the magnitude of the tibial slope has not been shown to consistently differ between dogs with and without CrCL rupture or between dogs with unilateral or bilateral rupture [32-34]. Various other anatomical features of the femur and tibia, including stenosis of the intercondylar notch, have been explored [28,35,36]; however, none have been consistently validated with repetition.

Two large retrospective studies reported age, neuter status and body weight to be risk factors for the development of CrCL rupture. Older dogs were reportedly at a higher risk [25] with a peak prevalence between 7 and 10 years of age. Neutered males and females were found to have a higher risk than intact males and females [26]. Higher body weight was found to increase the risk for rupture with an increased prevalence in dogs weighing >22kg [25]. Larger dogs were also noted to rupture their CrCL at a younger age. When these factors were examined with respect to likelihood of rupture of the contralateral limb in Labrador retrievers, age, weight and sex were not found to correlate with risk [33].

The pathologic changes present in ruptured cruciate ligaments have been described. Hayashi et al found decreased numbers of fibroblasts within the core of the CrCL with normal fibroblast density at the periphery [24]. The population of fibroblast subtypes was also altered with decreased numbers of fusiform and ovoid cells, and increased spheroid cells in the core. There was expansion of the epiligamentous regions with no evidence of scar formation indicating a lack of productive healing. Lastly there was loss of the crimping arrangement of the fibrils, an important biomechanical feature of the CrCL microstructure. An increased number of apoptotic cells has also been noted in diseased CrCLs [37]. The factors contributing to these degenerative changes, subsequent rupture and progression of osteoarthritis have also been investigated. Both inflammatory and immune-mediated causes have been proposed.

One study found increased synovial macrophage density in dogs with CrCL rupture [38]. The density of macrophages was correlated with severity of osteoarthritic changes and presence of interleukin-6 and tumor necrosis factor-α. These authors suggest that macrophages may play an βimportant role in the production of these destructive cytokines with subsequent progression of osteoarthritis. Other macrophage-associated factors that have been shown to have increased expression in the presence of CrCL rupture include tartrate-resistant acid phosphatase (TRAP), matrix metalloprotinease (MMP)-2, MMP-9, MMP-6, interleukin-1β, cathepsin S and cathepsin K [39,40]. TRAP and cathepsin K have been investigated to the greatest extent. Muir et al reported increased quantities of cathepsin K and TRAP within the ruptured CrCL [41]. These proteinases more heavily populated the epiligamentous tissue as compared to the central region. The author proposed that the collagenolytic activity of these proteinases contributes to the progression of degenerative changes within the stifle and the pathways leading to excessive activation of these proteinases may play important roles in the progressive degeneration that ultimately results in ligament failure. This same author further suggests in a more recent paper that it is the production of these proteinases by the synovium that may induce pathologic rupture of the cruciate ligament. This follow-up study documented the presence of increased cathepsin K and TRAP not only in the CrCL of dogs with rupture but also in the synovium; similar concentrations were not found in the intact CrCL [42,43].

In the autoimmune theory, antibodies develop against type-I collagen fibers in the CrCL in response to cyclical loading and microinjury possibly secondary to infiltration by synovial fluid [18] and to type-II collagen fibers in articular cartilage damaged by instability [44]. The specificity of these antibodies to CrCL disease however has not been documented. In one study, antibodies to collagen type-I and type-II were quantified in dogs with various joint diseases including complete CrCL rupture, partial CrCL rupture and osteoarthritis secondary to other underlying disease conditions [45]. While antibodies were elevated in the diseased state as compared to the controls, changes were not specific to the disease process. As evidenced by the multitude of studies investigating the predisposing, primary and perpetuating factors of CrCL disease without identification of a single most important factor, it is clear that much still has to be learned in regard to CrCL disease.

C. Surgical Techniques

Conservative management of rupture of the cranial cruciate ligament (CrCL) consists of weight management, exercise restriction/modification and anti-inflammatory medications as necessary. In dogs weighing <15kg, conservative management has a reported success rate of 86-90% [46,47]. The success rate in larger dogs is significantly less. In one study only 6 of 20 large dogs with cranial cruciate rupture became sound with conservative management [48]. In another study 81% of large breed dogs treated conservatively were still lame after 10 months [46]. In some of these cases the lameness progressed with time. As a result of these studies, surgical intervention is typically recommended for large dogs as well as small dogs that fail to respond to conservative management. The intent of surgical intervention is to provide stability to the stifle through restoration of normal joint mechanics, address concurrent meniscal injury and prevent the progression of osteoarthritis. Intracapsular and extracapsular techniques primarily aim to eliminate cranial tibial translation and stabilize the stifle to allow for periarticular fibrosis. Long term, this fibrosis becomes the primary restraint against cranial tibial translation [17]. Newer osteotomy techniques aim to dynamically stabilize the stifle during weight bearing without reliance on periarticular fibrosis [49].

1. Intracapsular techniques

The first technique reported for repair of the CrCL was the intra-articular procedure described by Paatsama [50]. This procedure attempts to replace the ruptured CrCL with a strip of fascia lata harvested from the lateral aspect of the stifle. The most distal aspect of the 1–2cm wide graft is the only portion that remains attached to the body. After the graft is harvested tunnels are drilled in the femur and tibia. The tunnel in the femur is drilled from the proximal aspect of the lateral collateral ligament to exit near the femoral origin of the CrCL. The tunnel in the tibia begins along the medial border of the tibial crest and extends proximally to exit at the tibial insertion of the CrCL. The graft is then pulled through the tibial and femoral tunnels. The graft is tightened with enough tension to eliminate cranial tibial translation and then sutured to the patellar insertion on the tibial tuberosity. This procedure is technically demanding, as it is very difficult to position the drill holes such that they exit exactly at the origin and insertion of the CrCL; the femoral tunnel is particularly challenging. Additionally, the fascia lata has significantly less tensile strength than an intact cruciate and as the graft exits the tunnels it is subjected to shear forces, which can further weaken the repair. In one study rupture of the graft was noted in 10 of 26 stifles in which it was performed [51]. However, only 4 of the 10 dogs were lame as result; periarticular fibrosis was likely the primary source of stabilization in those dogs where an appreciable lameness was not detected.

A modification of this technique was described by Dickinson and Nunamaker [52]. The fascia lata graft is harvested as described by Paatsama. However, a tunnel is only created in the femur and the graft is passed intra-articularly through this hole. It is secured to the edge of the fascia lata from which it was harvested. Dueland described a similar technique using the middle third of the patella and patellar tendon [53]. This graft is also passed intra-articularly through a tunnel in the femur and sutured to the soft tissues on the lateral aspect of the distal femur. Additionally the small patellar fragment is secured to the lateral femur with a small pin. Two cadaver limbs repaired using this technique were evaluated with respect to the instant center of motion, a parameter used to describe the normal motion at an articular surface [54]. In this study it was found that the patellar tendon graft did not disrupt the normal rolling motion of the articular surfaces, indicating that it preserves normal biomechanics throughout motion.

The over-the-top technique described by Arnoczky uses the medial third of the patella, patellar ligament and quadriceps tendon [55]. Unlike the other intracapsular techniques tunnels are not drilled in either the tibia or femur. Instead the graft, which remains attached to the tibial tuberosity, is pulled from cranial to caudal through the joint and then is passed from caudal to cranial over the top of the lateral femoral condyle. Hulse also described this same technique using the fascia lata, lateral retinacular fascia and lateral one third of the patellar ligament [56]. Use of the patellar ligament provides a graft of superior strength as compared to the fascia lata. However, this superior strength still does not approach that of the intact CrCL. An *in vitro* study comparing the biomechanical properties of the intact cruciate to a medial, lateral and central patellar tendon graft found that each graft only tolerated 10%, 29% and 29% of the load tolerated by the intact cruciate, respectively [57]. Similar results were reported in a study using research dogs [58]. Twenty-six weeks after repair of a surgically created CrCL-deficient stifle using the over-the-top technique, grafts only achieved 31% of the stiffness of a normal CrCL. This over-the-top technique has been shown to place the graft in an anatomically correct orientation as evaluated using the instant center of motion [59]. However, in a study assessing the clinical and radiographic outcome of this procedure, despite a decrease in the severity of lameness, all 21 dogs evaluated continued to have palpable instability, crepitus and joint swelling as well as progressive osteoarthritis when evaluated at a mean of 14.9 months after surgery [60].

The four-in-one over-the-top procedure is a modification of Arnoczky's procedure that combines both intracapsular and extracapsular repair techniques [17]. In this procedure a fascia lata graft is used in place of the patellar tendon in order to avoid the difficulties associated with harvesting the small wedge of the patella. The graft is passed from cranial to caudal as in the original over-the-top procedure. Additionally the caudal sartorius and biceps femoris muscles are advanced cranially and a lateral extra-capsular fabellar suture is placed. Finally, closure of the facial defect from which the graft was harvested imbricates the lateral retinaculum.

In the under-and-over technique described by Shires et al a fascia lata graft is harvested in the manner described by Paatsama along with the lateral one third of the patella ligament. The graft is then passed under the intermeniscal ligament, through the joint and up and over the top of the lateral femoral condyle [61]. The graft is then secured to the lateral femoral condyle with a spiked washer and screw. Clinical outcome in 38 stifles was reported in the initial description of this procedure [61]. By 3 months post-op 93% of the dogs showed significant improvement in lameness scores with no more than a slight limp noticed after exercise in some dogs. At final follow-up 95.7% of dogs had less than 1-3 mm of cranial caudal laxity, a significant improvement over pre-op values. However, radiographs showed progressive osteoarthritis of varying degrees in all but 2 stifles. A similar finding was reported in a subsequent study [62]. Despite owner satisfaction in >80% of these cases, another study showed significant decrease in limb function in dogs with experimentally transected CrCL treated with this procedure as evaluated using force plate analysis [63].

Various allogenic grafts and synthetic intra-articular prostheses have been described in the literature. Allografts have not gained popularity due to their tendency to incite an immune response even when stored at -70°C before implantation [64,65]. Synthetic materials including Dacron, Gore-Tex (W.L. Gore and Associates, Flagstaff, Az, USA) and carbon fiber have been reported [17]. However, the clinical application of these materials has not been extensively evaluated.

Despite the theoretical advantage of intracapsular techniques the failure to identify a graft material of sufficient strength has lead to the abandonment of these procedures by most surgeons. Much of the current literature focuses on extracapsular techniques. However, a recent report proposed a new graft with encouraging results. In a study by Lopez et al a hamstring graft was utilized [66]. The graft was composed of the combined tendinous insertion of the semitendinosus and gracilis muscles, the associated dense connective tissue and fascia from the medial aspect of the cranial tibial muscle. A modified over-the-top procedure was used in which the graft was passed through a tibial tunnel before being passed through the joint and over the top of the lateral femoral condyle. Two dogs were evaluated at 12 weeks and two at 52 weeks. At 12 weeks there was no evidence of osteophyte formation within the joint and grafts were highly vascular and undergoing active remodeling. By 52 weeks, lameness had completely resolved, there were radiographic changes consistent with only mild osteoarthritis, and the grafts were well vascularized and encapsulated by synovium. While these grafts were not biomechanically tested the encouraging clinical results suggest further testing may be warranted.

2. Extracapsular Techniques

The majority of extracapsular techniques involve imbrication of the lateral joint tissue with one or more sutures. The sutures are placed so as to prevent cranial tibial translation. However, by placing the constraint on the lateral aspect of the stifle the normal internal rotation that occurs with flexion is disrupted [54]. This alteration in normal biomechanics leads to abnormal motion at the articular surface [54]. Disruption of normal joint orientation and load is known to contribute to the progression of osteoarthritis [5]. As with the intracapsular techniques none of the extracapsular techniques appear to consistently reestablish normal biomechanics and fail to prevent the progression of osteoarthritis.

One of the first extracapsular techniques described the placement of multiple Lembert sutures in the periarticular tissues [67]. Three layers of sutures are placed. The first layer is placed at the junction of the fascia lata and the biceps femoris beginning approximately 2cm above the patella and extending to the level of the tibial tuberosity. A second layer consisting of 2-4 sutures is placed superficial to these sutures in the region between the patella and tibial tuberosity. A third set of sutures is placed on the medial aspect of the stifle in the periarticular tissues between the patella and tibial tuberosity.

Another early technique described the placement of one or two sutures on the lateral aspect of the stifle. After lateral arthrotomy heavy, nonabsorbable suture is passed around the lateral fabella and then through the lateral aspect of the distal patellar ligament forming a mattress pattern [68].

The two suture ends are tied to one another at the level of the fabella. In the report first describing this procedure, 42 stifles in 37 dogs were evaluated between 2 and 14 months post-op. Results were considered satisfactory, defined as no more than a slight lameness, in 87.7% of the stifles. This technique, termed lateral retinacular imbrication, became the basis for many of the subsequent techniques. This technique was evaluated using the instant center of motion [54]. It was found that unlike intracapsualr techniques, this extracapsular configuration did not preserve normal joint motion. This alteration resulted in the generation of a force that promoted joint compression and resisted the normal sliding motion at the articular surface. Additionally normal internal rotation was eliminated.

One of the first modifications of this technique described the placement of three lateral sutures [69]. All three sutures are anchored distally through the patella ligament. One suture is anchored proximally around the lateral fabella and the other sutures are placed through the lateral collateral ligament. In a study evaluating 34 stifles repaired via this modification 94.1% had no lameness at a mean follow up of 20 months [69]. Additionally 79.4% had no palpable drawer on physical exam. However, the majority of dogs had progressive osteoarthritis of the operated limb.

In the modification of the lateral retinacular imbrication technique described by Flo, a second or third suture is placed on the medial aspect of the stifle joint [70]. Additionally, instead of placing the suture through the patellar ligament a hole is drilled in the cranioproximal aspect of the tibial tuberosity through which both sutures are passed. In this technique the lateral suture is tightened first, followed by the medial suture. A third suture is also placed around the fabella and through the distal portion of the patellar ligament as in the early imbrication technique. These sutures function together to provide a “sling” to prevent cranial translation of the tibia. A slight modification of this technique, the three-in-one technique, also included the cranial advancement of the biceps femoris laterally and the caudal sartorius medially to provide extra support during the post-operative period [71]. The most popular extra-articular technique performed at this time is a simplified version of the Flo technique. In this technique two sutures are placed around the lateral fabella and passed through a tunnel in the tibial crest before being secured with a knot or crimp system. Complications of this technique were recently evaluated by Casale et al [72]. In this study 363 procedures were retrospectively evaluated. The overall complication rate was found to be 17.4% with 7.2% requiring second surgeries. While many factors were statistically evaluated only body weight and age were significantly correlated with complication rate. Heavier and younger dogs were found to be at a higher risk for complications post-op. Overall, the complication rate was similar to that reported for other procedures.

A unique technique also involving medial and lateral support, termed a posterolateral capsulorrhaphy, was described by Hohn [73]. This procedure involves the cranial transposition of the caudal sartorius muscle by suturing it to the craniomedial aspect of the patellar ligament. Laterally two mattress sutures are placed in the caudolateral joint capsule with the joint in flexion and externally rotated. A third suture is placed superficially around the lateral fabella and through the distal aspect of the lateral collateral ligament. Finally the biceps tendon and fascia lata are plicated over the patellar tendon. This technique effectively advances the active constraints of the stifle joint while simultaneously imbricating the caudolateral aspect of the joint.

An alternative extracapsular technique that has been evaluated is the fibular head transposition [74]. In this technique the fibular head, including the insertion of the lateral collateral ligament, is transposed cranially to function in a similar manner to the suture used in other techniques. The fibular head is secured to the tibia under the caudal edge of the cranial tibial muscle using a pin and tension band. Unlike the suture techniques, this procedure uses a vascularized graft. This is beneficial as the graft can undergo hypertrophy in response to the altered pattern of stress further strengthening the repair. Reported complications of this procedure include iatrogenic fracture of the fibular head, post-operative seroma formation, continued instability and progressive osteoarthritis [75]. One study compared immediate post operative cranial laxity and stiffness of four separate repair techniques: the over-the-top as described by Arnoczky, the under-and-over as described by Shires, Flo’s lateral retinacular technique with the exclusion of the medial fabellar-tibial suture, and the fibular head transpostion [76]. This study concluded that of the four techniques the fibular head transposition was biomechanically superior. This technique resulted in a significantly stiffer repair with less laxity then the other techniques. In the initial report describing this technique multiple evaluations of the technique were performed [74]. Four cadaver limbs were evaluated which demonstrated that the fibular head transposition normalizes craniocaudal instability, range of motion and inhibits normal internal rotation. Four research dogs were also evaluated 6 months post-op. Two of these dogs re-developed mild craniocaudal instability and the 3 dogs evaluated at 6 months also had increased internal rotation as compared to the post-op assessment. Radiographically these dogs did not show any evidence of degenerative joint disease 6 months after the procedure. The final group of dogs evaluated in this study included clinical cases. Seventy-one stifles were repaired using fibular head transposition while 25 were treated with the intracapsular fascia lata technique as described by Dickinson. Ninety percent of the fibular head transposition had a good to excellent outcome where-as 88% had a good to excellent outcome with the fascial repair as assessed by the owners. This difference was statistically significant with improved outcome in dogs treated with fibular head transposition. Higher body weights and concurrent meniscal injury were found to adversely affect outcome. In a similar study evaluating long term results in 80 dogs results were good to excellent in 90% of the dogs as evaluated by the owners [75]. However, a study assessing the outcome of 30 dogs treated using this technique reported concerning results [77]. All stifles developed cranial drawer motion and tibial rotation by 3 weeks post-op, both of which had been absent in all dogs immediately post-op. No dog had a full functional recovery in the limb as assessed by force plate analysis. By 4 months 25% had meniscal damage and by 10 months 50% had evidence of meniscal injury. All dogs had gross and histological degenerative changes within the articular cartilage. Thus, although this technique may be biomechanically superior, outcome does not consistently appear to be significantly improved as compared to other techniques.

3. Extracapsular Osteotomy Techniques

As both the intra and extracapsular repair methods have resulted in suboptimal stabilization of the cranial cruciate ligament (CrCL)-deficient stifle, newer procedures have been designed in an attempt to alter the biomechanics of the stifle to eliminate the necessity of an intact CrCL for prevention of cranial tibial translation. Using some form of tibial osteotomy these procedures alter the biomechanics of the stifle to provide functional craniocaudal stability during weight bearing. At the midpoint of the stance phase at a walk the stifle joint is held at approximately 135° [78]. Due to the anatomy of the femorotibial joint surface a cranially directed shear force is created between the two bones during loading. This femorotibial shear force is opposed by the CrCL in a normal stifle. In a CrCL-deficient stifle this force is unapposed resulting in cranial translation of the tibia. Functional neutralization of this femorotibial shear force is the goal of these tibial osteotomy procedures. This approach to cranial cruciate disease was first proposed by Slocum in 1984 who described the cranial tibial closing wedge osteotomy (CTWO) [79]. This procedure involves removing a triangular piece of bone with its base on the cranial aspect of the tibia and then securing the osteotomy with a bone plate. The osteotomy is made as proximally as feasible to still allow for placement of at least 3 screws in the proximal segment. The wedge angle to be excised would intuitively be equal to the angle of the tibial plateau minus the desired post-operative angle of 5°. However, this procedure results in a cranial shift of the functional axis of the tibia leading to inadequate leveling of the tibial plateau. As a result, "over rotation" of the tibial plateau is necessary to achieve functional stability using this procedure. This variability in achieving the appropriate post-operative angle was demonstrated in the study by Macias et al. Post operative angles ranged from 7° to 20° [80]. A few case series have reported clinical outcome after the CTWO. Slocum reported this technique in combination with cranial advancement of the semitendinosus, gracilis and biceps femoris muscles in a series of 19 stifles [79]. These dogs had a rapid recovery with full return to function in 9 of 9 dogs evaluated long term. Cranial drawer could not be palpated in any of the stifles. Healing of the osteotomies was uncomplicated and subjective evaluation of follow-up radiographs did not identify progression of osteoarthritis in 15 of the 19 cases. Another retrospective study of 91 dogs identified a similar success rate with 86% having good-to-excellent limb function based on owner assessment [81]. Two additional small case series that used the CTWO in dogs with CrCL rupture and proximal tibial deformities reported similar results [80,82]. Complications associated with this procedure primarily include implant failure, tibial fracture and nonunion [79,82]. In one case series an 11.9% re-operation rate was reported, primarily for catastrophic implant failure[83].

In 1993 Slocum proposed the tibial plateau leveling osteotomy (TPLO) [84]. The biomechanical properties and clinical outcome of the TPLO have been extensively described in the literature. These studies will be discussed in detail in the following section.

In the TPLO rotation of the proximal fragment isolates the proximal portion of the tibial tuberosity. As the powerful quadriceps muscle inserts on the tibial tuberosity it has been suggested that loss of buttressing normally provide by the proximal fragment may predispose the tibial tuberosity to fracture. This is of particular concern in dogs with large tibial plateau angles (>34°) that require significant rotation of the proximal fragment to achieve a post-operative angle of 5°. To minimize this loss of buttressing while still achieving the desired post-op TPA combination of the TPLO and CTWO has been proposed. This combination was first described by Talaat et al in a case series of 15 dogs [85]. Pre-operative TPAs ranged from 35° to 53°. In the first cases the TPLO was performed, the proximal segment was rotated 12 mm (TPA 34°) and the remainder of the necessary rotation was achieved using the CTWO. For simplicity, the CTWO was later standardized to 10° or 15° and the remainder of the angle was addressed with the TPLO. Additionally, in the first 4 cases the TPLO was performed and the biradial saw was used to create the CTWO. However, as this resulted in a mean post-operative TPA of 16° the CTWO was subsequently performed with a sagittal saw, which improved the mean post-operative TPA to 5°. Fixation of the three fragments was achieved using a TPLO plate and either a limited contact dynamic compression plate (LC-DCP) or a pin and tension band. While final outcome in these dogs was good to excellent in 90% of the dogs at a mean follow-up of 23 weeks with all owners being satisfied with the outcome, an impressively high complication rate, 77.8%, was reported. Complications including patellar tendon thickening, implant loosening or breakage, seroma formation and local irritation were reported. One third of the cases required a second surgical procedure to address the complication. This trend in owner satisfaction in the face of a significant complication rate was reported in another study surveying techniques used by surgeons for excessive tibial plateau [86]. In this study six dogs were treated with TPLO plus CTWO for excessive TPA. All dogs had complications including patellar ligament thickening, loss of TPA and tibial tuberosity fracture. Sixty-seven percent of these required a revision surgery. Despite this high complication rate, when the overall owner assessed outcome was compared for dogs treated for excessive TPA to those with normal TPA there was no significant difference. However, the combination of TPLO plus CTWO was not differentiated from other techniques used for excessive TPA including TPLO alone, CTWO alone and TPLO plus a lateral fabellar suture.

Another tibial osteotomy technique described in the literature to a limited extent is the proximal tibial intraarticular osteotomy (PTIO) [87,88]. This procedure levels the tibial plateau by removing a 10° to 25° wedge of bone based at the articular surface of the proximal tibia. The base of the wedge is defined by the bursa of the patellar ligament cranially and the cranial aspect of the medial meniscus caudally. The apex extends to the caudodistal aspect of the medial collateral ligament. In the original description the osteotomy was secured using two craniocaudal positional screws [87]. Additional support was provided by a medial and lateral fascial imbrication. Outcome using this technique was described in 100 dogs. Intra-operative complication rate was 19% including 12% valgus malalignment, 2% intra-articular screw placement, 1% popliteal artery laceration, 1% cranial tibial artery laceration, 1% peroneal nerve injury, and 2% long digital extensor tendon injury. The majority of these complications required further surgical procedures. The post-operative complication rate was 12% including proximal tibial fracture, subsequent meniscal injury, and decreased range of motion due to fibrosis around the long digital extensor muscle. All required second surgical procedures. At four months post-op 75/87 dogs with complete CrCL rupture were completely sound, 5/87 were mildly lame and 7/87 were moderately lame. All of the 18 dogs with partial CrCL rupture were clinically sound and none had subsequent meniscal injuries. Despite the final outcome the authors in this study did not recommend this procedure as a valid alternative to the TPLO due to the high complication rate.

A subsequent report modified this technique in attempt to minimize the complication rate [88]. This new technique included a caudal medial meniscal release, osteotomy of the fibula and fixation using a medial bone plate in addition to a single craniocaudal positional screw. The fascial imbrication was not included in this description. This technique was reported in 60 stifles [88]. A 13% intraoperative and 7% post-operative complication rate was reported. Unlike the previous report no dogs required a second surgical procedure to address the complication. Only 1 dog that became acutely lame 9 months after surgery required implant removal for possible osteomyelitis. At 12 months after surgery 91% were completely sound. There was no significant change in range of motion from preoperative values nor were any subsequent mensical injuries reported; however significant progression of osteoarthritis did occur. With the improved complication rate and high long-term success, these authors concluded that PTIO is an acceptable alternative to the TPLO.

One of the newest osteotomy techniques is the tibial tuberosity advancement (TTA). This technique neutralizes cranial tibial thrust through advancement of the patellar tendon such that it lies perpendicular to the tibial plateau. This is achieved by making a longitudinal osteotomy of the tibial tuberosity [89]. The fragment is then advanced cranially, secured in this position using a 3, 6, 9, or 12mm cage and then stabilized using a specialized tension-band bone plate on the medial aspect of the tibia. Finally bone graft is placed in the defect to help promote healing. The biomechanical success of this procedure has been reported in multiple studies [90-92]. A new study evaluated the three-dimensional effects of the TTA on stifle kinematics [93]. This study by Kim et al not only demonstrated the elimination of cranial tibial translation after repair using the TTA but also found that the TTA normalized internal rotation as well as femorotibial contact mechanics. Despite these biomechanical advantages, like the other osteotomy techniques, complications are reported in a significant number of cases. In the first study documenting clinical outcome using this technique in 65 stifles an overall complication rate of 59% was reported [94]. While the majority of these complications consisted of incisional swelling and bruising, 21% were considered more serious, including implant failure, tibial crest fracture/displacement and subsequent meniscal injury. Additionally 67% showed radiographic progression of osteoarthritis during follow-up. However, despite complications and progressive osteoarthritis, a good to excellent outcome was reported in 90% of the cases as assessed by owner questionnaires. A subsequent study reported similar results for overall outcome with a significant decrease in complications [95]. In this study by Lafaver et al nearly 100% of the owners were satisfied with the procedure with 83.1% reporting a good to excellent outcome. Complications occurred in 31.5% with only 12% being classified as major complications, as defined by the need for a second surgery. Major complications included tibial fracture, implant failure, subsequent meniscal injury, infection, lick granuloma, incisional trauma and medial patellar luxation. All major complications were successfully treated and at the time of final in-hospital reevaluation 74.5% were free of lameness, 23.5% had a mild lameness and 3% had a moderate to severe lameness. A third study objectively evaluated post-procedural outcome using force plate analysis [96]. While all but one of 40 stifles had improved ground reaction forces, the values were still significantly lower than control dogs; most dogs only reached approximately 90% of the values recorded in control dogs at last follow-up (mean 5.9 months; range 4 – 16 months). This study also assessed the effect of breed, body weight, meniscal injury, degree of CrCL tear, articular cartilage damage and concurrent orthopedic injury on follow-up peak vertical force and found no relationship. The only factor found to influence outcome was the presence of complications with those dogs having lower peak vertical forces at follow-up. This study reported a complication rate similar to that of Lafaver (25% with a 12.5% re-operation rate).

A similar procedure that does not require the specialized implants used in the TTA has also been described, the triple tibial osteotomy (TTO) [97]. In this procedure an incomplete longitudinal osteotomy is made isolating the tibial crest; the most distal portion of the fragment remains attached to the tibia. A partial wedge ostectomy based at the cranial aspect of the major tibial segment is then made in the proximal tibia. The size of the wedge is approximately two-thirds of the angle required to achieve a post-operative patellar tendon – tibial plateau angle of 90°. The wedge is then reduced and the proximal and distal fragments are secured with a bone plate. Closure of the wedge simultaneously advances the tibial tuberosity and reduces the tibial plateau angle. Only two studies have documented the clinical outcome of this technique. In the first study, which originally described the procedure, an overall complication rate of 36% (23/64 stifles) was reported with the most common complication being fracture through the distal end of the tibial crest during tibial crest advancement. This fracture was repaired intra-operatively using a pin and tension-band. Complications occurred in 10% post-operatively with 6% requiring revision surgery for subsequent meniscal injury, infection or tibial tuberosity fracture. At long-term follow-up ranging from 11 to 26 months there was a significant increase in thigh circumference and stifle range of motion as compared to pre-operative values. No radiographic progression of osteoarthritis was noted. However, cranial drawer was present in all cases and the tibial compression test was positive in >90%. Despite the continued presence of palpable instability, 100% of the owners reported a marked improvement in their dog's quality of life after the procedure and would choose to have the procedure performed again. The second study evaluating this technique reported an overall post-operative complication rate of 23% (22/97 stifles) with the type of complications being much more varied than in the original study. Complications included tibial crest avulsion, fracture of the distal tibial attachment of the tibial crest, fibular fracture, patellar tendonitis, subsequent meniscal injury, implant failure and patellar fracture. Additionally increased patient age, increased wedge angle and intra-operative fracturing of the cranial tibial cortex were found to be associated with post-operative avulsion of the tibial crest. Despite a comparable complication rate, the limited evaluation of this procedure thus far makes comparisons to other surgical techniques difficult.

D. Tibial Plateau Leveling Osteotomy

1. Surgical technique

The tibial plateau leveling osteotomy (TPLO) procedure involves making a radial osteotomy in the caudoproximal tibia and rotating the proximal segment to achieve a post-operative tibial plateau angle (TPA) of approximately 5°. Prior to surgery the TPA must be determined from lateral radiographs [98]. Proper radiographic technique with the beam centered over the stifle and superimposition of the femoral and tibial condyles is vital to assure accurate measurement. Deviation from this positioning has been shown to over or underestimate the TPA with cranial or proximal positioning and caudal or distal positioning, respectively [99]. The most widely accepted technique for determination of the TPA is based on the identification of specific landmarks. On a lateral radiograph a line is drawn along the tibial plateau from the most cranial point of the medial tibial plateau to the most caudal point, corresponding to the tibial insertions of the cranial and caudal cruciate ligaments, respectively. Another line is drawn along the functional long axis of the tibia from the intersection of the medial and lateral tibial intercondylar eminences proximally to the center of the talus distally. A line is then drawn perpendicular to the long axis of the tibia intersecting the intercondylar eminences. The angle formed between this perpendicular line and the line along the tibial plateau is the TPA [98]. While other methods of determining the TPA have been described [9,100,101], this is currently the most widely published technique. Various studies have been performed comparing measurement of TPA using conventional and digital techniques. These studies have shown that anatomically accurate measurements are equivalent [102] if not superior using the digital imaging [101,103] as compared to conventional images. Intra- and inter-observer variations using this technique have been reported to range from 1.5 to 3.4° and 0.8 to 4.8°, respectively[104,105]. One study reported the consequence of this inter-observer variability to be minimal resulting in < 1mm of variation in rotational values [102]. While experience has been reported as influential in inter-observer variation in normal dogs [105], a study in dogs with CrCL rupture did not support this finding [104]. This study instead found that degenerative changes, particularly along the caudal aspect of the tibial plateau, were most influential in variability of TPA measurements [104]. To avoid the effect of degenerative changes on accurate TPA an extension technique to identify the caudal point on the tibial plateau or use of the opposite limb for TPA measurement have been proposed and validated [106].

The TPLO procedure begins with a standard medial approach to the stifle [107]. After the initial incision the pes anserinus (semitendinousus, gracilis and caudal belly of the sartorius) and the associated fascia are elevated to expose the cranial proximal aspect of the tibia. In the original report a complete craniomedial arthrotomy is then performed to inspect the CrCL, debride the remnants, evaluate the medial meniscus and perform an axial meniscal release of the caudal horn. While some surgeons still employ this technique multiple variations have been proposed and debated without a clear benefit. The necessity of debridement of the remnants of the CrCL has been debated [108]. The risk versus benefit of a full arthrotomy has also been questioned with some surgeons opting for a mini-arthrotomy or arthroscopic evaluation of the stifle anatomy. A limited caudal arthrotomy has been reported to significantly lessen the progression of osteoarthritis in TPLO repaired stifles as compared to an open medial arthrotomy [109]. Arthroscopic evaluation is reported to decrease post-operative morbidity when objectively compared to open arthrotomy using force plate analysis [110]. Additionally, a multitude of studies have been performed in attempt to evaluate the necessity and consequence of performing a meniscal release without a clear consensus [111-116].

After addressing the intracapsular components preparation of the tibia is completed through subperiosteal elevation of the popliteus muscle. Two gauze sponges are then placed between the lateral musculature and vasculature to protect them from damage by the saw. The necessity of these sponges was recently evaluated in a cadaveric study [117]. This study found that use of the gauze sponges only improved the technique via protection of the caudoproximal tibial musculature; no damage to the cranial tibial muscle or the popliteal artery was noted with or without the sponges. The authors concluded that manual retraction of the caudoproximal tibial musculature would prevent damage without the need to disrupt the lateral musculature with gauze sponges.

The next step is the placement of a custom designed jig. Two partially threaded pins are placed in the tibia at the medial collateral ligament and the distal diaphysis. These pins are used to secure the custom jig to help maintain alignment of the proximal and distal segments during fragment rotation. While in the original description the jig is recommended recent reports have shown that the procedure can be performed successfully with or without the jig [118,119].

Following jig placement a biradial saw blade is used to create the caudoproximal osteotomy. The saw is positioned perpendicular to the medial aspect of the tibia in both the cranial-caudal and proximal-distal directions. The caudal aspect of the blade should exit the caudal aspect of the tibia perpendicular to the cortex. Blade size is chosen based on body weight and tibial dimensions. Blades are available in 12, 18, 21, 24, 27 and 30mm. Prior to completion of the osteotomy through the lateral cortex reference marks are placed on the fragments such that rotation of the fragments aligns these marks achieving a TPA of approximately 5°. A TPA of 5° allows for the elimination of cranial tibial thrust without placing excessive strain on the caudal cruciate ligament [10] . The magnitude of rotation (C) is calculated from the original tibial plateau slope (α) and the radius of the osteotomy (R) using the formula: $C = 2R(\sin [\alpha/2])$ [98]. The osteotomy is then completed and the proximal fragment is rotated caudally, aligning the marks, and then temporarily secured using a small pin through the proximal aspect of the tibial tuberosity. Rotation of the fragment can result in a stairstep between the medial cortices of the fragments; no effort should be made to align the medial cortex as this has been reported to adversely affect limb alignment [120].

Permanent fixation of the fragments is then achieved using a specialized TPLO plate that allows for a minimum of 6 cortices to be engaged in both the proximal and distal fragments. After plate application all stabilizing pins are removed and the procedure concludes with standard soft tissue closure. Many different orthopedic implant manufactures produce TPLO plates that vary slightly in composition and configuration. The original plate produced by Slocum Enterprises has been criticized due to concerns of implant corrosion being linked to neoplastic transformation of the underlying bone [121]. The presence of implant corrosion and bi-product tissue contamination has been documented in two separate studies [122,123]. These studies microscopically examined the implants and found that despite strict processing guidelines for casting of these plates the implants were often sub-optimally composed resulting in corrosion of the implant. This corrosion is thought to cause a predisposition to tumor development. Thus, use of this specific plate has been replaced by other designs. Newer plates are manufactured from wrought alloy instead of cast in hopes of avoiding the detrimental effects of corrosion.

Various TPLO plate constructs have been biomechanically compared. One study compared the stiffness of three commonly manufactured TPLO plates: Slocum, Securos and Synthes plates [124]. The study found the Slocum and Securos plates to have similar stiffness. The Synthes plate was significantly stiffer than the Slocum plate. Modes of failure for the similarly stiff Slocum and Securos plates were confined to plastic deformation. The majority of the Synthes plates underwent elastic deformation. A similar subsequent study using a gap model evaluated the properties of other TPLO plate designs [125]. No significant differences were noted in stiffness during axial compression when the Slocum, Securos, New Generation locking plate and New Generation broad locking plates were compared. However, the mean loads at failure were higher for with the locking plate designs. In a study by Leitner et al differences between fragment fixation using conventional and locking screws were mechanically compared [126]. Despite small increases in rotation and translation of the proximal fragment during plate application with the conventional screws, there was no difference in stiffness or cycles to failure between the two repair techniques. These authors suggest that careful contouring of the plate prior to fixation using conventional screws is as biomechanically effective as using locking screws.

2. Biomechanics

Various biomechanical aspects of the TPLO have been investigated including cranial tibial translation, internal rotation and variations in surgical technique.

In the active model of stifle stability proposed by Slocum equilibrium within the stifle joint is dependent on both active and passive forces [84]. Throughout motion there is a balance of moments around the instant center of motion provided by the active forces produced by muscle contraction and relaxation and the passive restraints of the CrCL and, to a lesser extent, the caudal horn of the medial meniscus. The active forces are the primary restraint and the passive forces are only recruited when the active forces do not sufficiently counteract cranial tibial translation. These mechanisms counteract the force, or cranial tibial thrust, produced by the caudal slope of the tibial plateau and compression of the articular surfaces during weight bearing. In the absence of a functional CrCL the active components are not able to counteract this cranially directed force effectively resulting in cranial tibial translation. As the only component of cranial tibial thrust available for manipulation is the slope of the tibial plateau, the cranial closing wedge osteotomy was proposed as a method to provide functional stability of the knee [79]. By decreasing the slope of the tibial plateau the point of contact between the articular surfaces moves from a cranial to caudal position to lie on the functional axis of the tibia resulting in the generation of a purely compressive force during loading. Thus there is no need for restraint against cranial tibial thrust, as it, theoretically, is no longer produced with loading. While the initial report describing the cranial closing wedge osteotomy had promising results [79] the procedure was subsequently modified such that leveling of the tibial plateau was achieved using a radial osteotomy of the proximocaudal tibia, or a TPLO.

Elimination of cranial tibial thrust and thus cranial tibial translation by the TPLO has been demonstrated in multiple biomechanical studies. Warzee et al showed that 18.9 ± 3.4mm of tibial translation generated by CrCL transection was converted to 3.2 ± 0.8mm at a standing angle of 135° after repair using the TPLO [10]. Kim et al found that 15.3mm of cranial tibial translation was converted to 2mm of caudal tibial translation post-TPLO at 135° [12]. Reif et al found that 12 ± 2.3mm of cranial tibial translation was converted to 2 ± 2.9mm of caudal translation at 120° of extension [11]. This study also found a linear correlation between magnitude of caudal tibial thrust and load. With increasing axial loads caudal tibial thrust values also increased [11]. This caudal tibial translation has been challenged in a biomechanical model proposed by Shahar et al [127]. This study found that the cranial directed force generated by weight bearing was not eliminated with a TPLO rotated to a post-operative angle of 5°. Only after rotation to 0° did minimal force remain in the cranial direction with the primary force directed caudally.

However, as this mathematical model represented a TPLO repaired stifle with an intact CrCL the conclusions of this paper can be questioned.

Only two biomechanical studies have evaluated the effect of the TPLO on internal rotation. In one study 23.3 ± 6.8° of internal rotation was attenuated to 9.5 ± 6.3° but not eliminated after TPLO rotated to 0° [10]. This result was justified by the "screw home mechanism" [17]. As the stifle is flexed, the lateral collateral ligament becomes lax while the medial collateral ligament remains taut, resulting in a progressive allowance for greater internal rotation with flexion. With caudal rotation of the tibial plateau the articular surface is positioned in a relatively greater degree of flexion [98]. This would, in turn, allow for more internal rotation at a normal standing angle post TPLO. However, another study did not support this finding. In a study by Kim et al 14.3 ± 2.9° of internal rotation was reduced to 1.2° of relative external rotation after repair with TPLO rotated to 5° [12]. This value was not significantly different from the intact values suggesting the TPLO effectively normalizes internal rotation at a standing angle of 135°.

Surgical technique has been shown to influence the generation of cranial tibial thrust. In the original description by Slocum the osteotomy was to be centered at the center of motion of the stifle, a point overlying the distal femur [84]. However, this proximal location made placement of the compression plate difficult. Centering the osteotomy 5mm distal to the tibial plateau through the medial collateral ligament was then proposed [10]. Mathematical evaluation of the most biomechanically appropriate osteotomy positioning was subsequently performed [128]. In this study by Kowaleski et al centering the osteotomy at the intercondylar eminences maintained the normal long axis of the tibia. Deviation from this position resulted in movement of the intercondylar tubercles and a shift in the long axis of the tibia resulting in unexpected post-operative tibial plateau angles. A subsequent biomechanical study showed that distal positioning of the osteotomy resulted in the continued presence of cranial tibial translation, a post-operative tibial plateau angle of 7.92 ± 0.49° and a long axis shift of 3.25 ± 0.27° [129]. By comparison centering the osteotomy over the intercondylar tubercles resulted in a post-operative tibial plateau angle of 5.50 ± 0.32° and an insignificant long axis shift. Based on these results a proximal osteotomy centered over the intercondylar tubercles is recommended to optimize postoperative tibial plateau angle with subsequent elimination of cranial tibial translation.

Orientation of the osteotomy has been shown to affect post-operative tibial angulation and torsion. In one study multiple variations in osteotomy angles and tibial plateau rotation angles were evaluated in a bone model [120]. Results showed that deviation in any direction from an osteotomy made perpendicular to the craniocaudal and proximodistal axes resulted in greater angular and rotational deformities. Additionally, greater tibial plateau rotations, ranging from 0 to 30°, created greater deformations. However, osteotomy reduction, either completely reduced or aligned along the medial cortex, was found to have a greater effect on deformities. Those with complete reduction had less angular and torsional deviation. These authors concluded that to decrease iatrogenic deformation the osteotomy be made perpendicular to both the proximodistal and craniocaudal axes and the osteotomy should be fully reduced without regard to medial cortex alignment.

In a biomechanical study by Windolf et al the effect of increasing tibial plateau rotations on stability was evaluated [130]. In this study a linear correlation was found between increased rotations and increased amounts of plastic deformation during cyclical loading. The authors suggested that dogs with greater TPAs may be predisposed to biomechanical failure with higher degrees of rotation.

3. Complications

In the initial tibial plateau leveling procedure described by Slocum, cranial closing wedge osteotomy was performed in 19 stifles [79]. Excellent outcome was reported in 9/9 dogs followed out to 1-year. Additionally, no cranial drawer motion was palpable nor was any progression of osteoarthritis noted on radiographs. Similar results were reported in a case series of 394 stifles repaired using the cranial closing wedge technique or TPLO [84]. Good to excellent functional results were reported in 94% of cases followed long term. Radiographic findings were not reported in this series; however, lack of progression of osteoarthritis was implied with the 5 descriptors of success: 1) return of full stifle flexion, 2) return of full hindlimb musculature, 3) a quiescent joint, 4) lack of radiographic progression of osteoarthritis, and 5) return to full function. Complications in this series included implant failure, subsequent meniscal injury, patellar luxation and rotary instability with a 8.4% re-operation rate for complications [84]. While these initial descriptions boasted the benefits of the TPLO over intra-capsular and other extra-capsular techniques, subsequent studies have debated this claim.

A study in 2004 reported radiographic findings in 40 dogs 6 months after being treated with TPLO. In this series 40% of dogs showed progressive osteoarthritic changes, 57.5% showed no progression and 2.5% showed a decreased in osteophytosis [131]. In a study by Lazar et al radiographic osteoarthritis scores were compared pre and >12 months post-operatively in 66 dogs treated with extracapsular repair or TPLO [8]. While dogs with larger scores were approximately 6 times more likely to have had an extracapsular repair, significant progression of osteoarthritis was noted with both techniques. Similarly, in a study comparing radiographic scores between dogs treated with TPLO and either a medial parapatellar arthrotomy or a limited caudal medial arthrotomy, all 68 dogs had radiographic progression of osteoarthritis with a follow-up ranging between 7 and 38 months [109]. In a larger study evaluating short-term progression of osteoarthritis a small but significant increase in radiographic scores was noted at 8 weeks after surgery [132]. However, 22.7% of dogs had a lower score at the time of follow-up. In a recent abstract comparing dogs receiving a lateral fabellar suture or a TPLO, radiographic progression was noted in both treatment groups evaluated at 24 months post-operatively [133]. Thus, although not absolute, the majority of dogs treated with TPLO have progression of osteoarthritis despite this being initially proposed as a benefit of the TPLO over other techniques.

A similar lack of superiority of the TPLO over other techniques has been reported in multiple studies with respect to functional outcome. In a study by Conzemius et al outcome was compared using force plate analysis in Labradors receiving a lateral suture, an intracapsular repair or a TPLO [6]. While dogs treated with intracapsular techniques had significantly less improvement (only 15% improved), there was no difference between those dogs treated using a lateral suture or TPLO. Improvement was only noted in 40% of dogs treated using a lateral suture and 34% of dogs treated using the

TPLO evaluated 6 months after surgery. However, increased peak vertical force and vertical impulse were noted in all dogs treated with a lateral suture or TPLO at 2 and 6 months after surgery. This study contradicted an earlier study in normal dogs treated with a TPLO [134]. In that study 6 normal dogs were treated with a TPLO and evaluated using force plate analysis at 8 and 18 weeks after the procedure. While peak vertical forces and vertical impulses were decreased at 8 weeks, values had normalized by 18 weeks. However, comparison of this study with the previous study is difficult due to the use of normal dogs. A systematic review of the literature using evidence based medicine failed to identify a superior surgical technique for repair of the CCL-deficient stifle [7]. Likewise, a study comparing the outcome of TPLO and cranial closing wedge osteotomy based on a client questionnaire did not identify a significant difference in outcome between groups with 92.3% reporting a good to excellent outcome [135]. Two recent abstracts also failed to identify a difference in outcome when comparing the lateral fabellar suture and TPLO. In one study, all dogs showed functional improvement and there was no significant difference between peak vertical force, thigh circumference or range of motion in either group at any time point up to 24 months after surgery [7]. In the second study no difference was found between the two procedures with respect to time to and completeness of recovery. These dogs were evaluated using force plate analysis up to 1 year after surgery with dogs in both groups achieving normal limb loading [133].

While differences in range of motion were not detected between the lateral suture and TPLO in this study by Au et al [133], decreased range of motion has been reported as a factor influencing the outcome post-TPLO. In a study by Jandi et al range of motion was evaluated in 412 stifles [136]. Three hundred and twenty two dogs had no loss of flexion or extension, 78 had <10° of flexion or extension loss and 12 had >10° of flexion or extension loss. Those dogs with loss of >10° of motion were significantly more likely to have a higher clinical lameness score. Loss of extension was found to be less tolerable and less responsive to aggressive physical therapy. Another study also reported failure of range of motion to normalize after TPLO; however, no correlation between continued lameness and range of motion was identified [137].

A complication rate ranging from 18.8% to 33.3% has been reported in association with the TPLO [135,138-140]. Complications include tibial crest fracture, fibular fracture, implant failure, joint infection, incisional infection, incisional edema, osteomyelitis, hemorrhage, meniscal injury, luxation of the long digital extensor tendon [141], peri-implant sarcoma formation [121,142], distal patellar tendon thickening [143] and patellar desmitis [143,144]. Risk factors have been identified for a number of these complications. Simulanteous bilateral TPLO procedures, loss of TPA post-operatively, a thin tibial crest post-osteotomy, age, weight and location of the anti-rotational pin have been shown to predispose dogs to tibial crest fractures [120,145,146]. Higher body weight, magnitude of tibial plateau rotation and pre-operative TPA have been shown to increase the risk of fibular facture [147]. Additionally, procedures performed without a jig are reported to have an increased risk for fibular fracture [147]. Risk factors for patellar ligament thickening include a cranial osteotomy, a partially ruptured CrCL in combination with a cranial osteotomy and post-operative tibial tuberosity fracture [143]. For the more severe patellar tendon condition, patellar desmitis, risk factors include a partially intact CrCL, higher body weight and a post operative angle of less than 6° [143,144]. Additionally one study reported an increased risk for complication development in Rottweilers and dogs receiving medial arthrotomies, but a decreased risk for Labradors [140]. This study also noted that the incidence of major post-operative complications was similar to values reported for other surgical techniques for repair of the CrCL-deficient stifle.

Loss of tibial plateau rotation has been noted with healing of the osteotomy [106,148]. This increase in TPA could potentially lead to return of cranial tibial translation. However, a clinical consequence to this finding has not been recognized. Additionally a study comparing post-operative TPA and ground reaction forces in Labradors found no difference in outcome with TPAs ranging from 0 to 14° [149].

E. Kinematics

Kinematics describe the temporal and geometric characteristics of motion [150]. Kinematics have been used to descriptively evaluate gait in the dog. Two or three-dimensional evaluation of gait is performed by placing markers over specific anatomical landmarks and then walking or trotting the dog through a series of cameras. The video is then analyzed using a computer, which then describes the character of each joint throughout the entire range of motion. Multiple kinematic studies have been performed in normal dogs. Hottinger et al described the gait of healthy large breed dogs at a walk [78]. In the hindlimb motion of the coxofemoral joint was characterized by a period of slow extension during the early stance phase followed by continued extension during the remainder of stance followed by flexion through the swing phase. Joint angles ranged between approximately 100 and 140° of flexion during the stance phase and 90 to 130° during the swing phase. Motion in the stifle joint was described by gradual flexion throughout the stance phase with two periods of slowed flexion during the early and late stages. Rapid flexion was then noted during the late stage of stance followed by rapid extension during the latter 2/3 of the swing phase. Stifle angles ranged from approximately 120 to 145° of flexion during the stance phase and 110 to 150° during the swing phase. The tibiotarsal joint was defined by initial flexion with impact followed by extension during the majority of stance with rapid flexion at the end of stance. The swing phase was composed of continued flexion followed by rapid extension during midswing. Flexion was then initiated in late swing that prepared the joint for impact. Joint angles ranged from approximately 130 to 170° during stance and 120 to 150° during swing.

Two studies have kinematically described hindlimb gait at a trot. In the first study using greyhounds the stance phase of the coxofemoral joint was characterized by progressive extension with a single peak at the end of stance phase [151]. During the swing phase there was flexion followed by gradual extension in preparation for impact. Mean joint angles ranged from 110 to 135° in stance and 100 to 135° during swing. In the stifle joint there were two peaks of extension, one at the end of swing phase and one of decreased magnitude at the nearing the end of stance. Motion was limited between 130 and 140° during stance but ranged from 95 to 150° during swing. Motion in the tibiotarsal joint consisted of gradual flexion at the beginning of the swing phase followed by extension, peaking just before the beginning of stance. The stance phase began with a short period of flexion followed by extension, peaking at a value similar to that of the swing phase. During stance the mean joint angle ranged from approximately 130 to 155° and 120 to 155° during swing. In a follow up study using mixed breed dogs, similar patterns of gait were identified in the hindlimb [152]. Excursional variations tended to be slightly larger than in the uniform population of dogs but the overall variance remained low.

When kinematically comparing the patterns of motion of the walk and trot two primary differences have been noted [78]. During the stance phase of the walk there tends to be more points of inflection indicating a more complete gait pattern during a walk as compared to the trot. This difference, however, was not noted in another study [2]. Additionally the angles of excursion noted at the walk tend to be less than those noted at a trot [78].

While these studies characterize the normal patterns of gait in the hindlimb, multiple studies have been performed that describe alterations in gait after CrCL rupture. In a study by DeCamp et al, patterns of gait at a trot were compared in dogs prior to and at 1, 3 and 6 months after CrCL transection [153]. As compared to the intact condition the stifle was carried in a more flexed position throughout the stance phase and early swing phase. This adaptive change corresponded to a reduced contribution of the stifle to propulsion. In contrast the coxofemoral and tibiotarsal joint tended to be more extended during the stance phase likely as compensation for the increased flexion of the stifle joint. This adaptive change in stifle flexion was also noted in another study in which a 5-14° increase in flexion was noted in dogs at a walk 7 weeks after CrCL transection [2]. The magnitude of this change however was less pronounced at a trot. A third study evaluating changes in gait at a walk up to 24 months post-transection also noted a similar increase in flexion but changes were only significant during the early (i.e.< 6mo) period [3].

Two of these studies also documented changes in internal rotation and cranial tibial translation during gait in the CrCL-deficient stifle. In the study by Korvick et al, 7 weeks after transection internal rotation was increased in dogs at a trot [2]. This change was most pronounced during the mid-stance phase. As the screw home mechanism in the stifle joint allows for greater degrees of internal rotation with increased flexion this finding is consistent with the stifle being carried in a more flexed position after transection [17]. However, this finding is contradicted in the study by Tashman et al that found a non-significant trend towards reduced internal rotation up to 24 months after transection in dogs evaluated at a walk [3]. These authors suggest insufficient torque is generated during a walk to produce significant internal rotation. They further suggest that anatomical factors other than the CrCL may play a greater role in rotational generation and restriction. These two studies are more in agreement with respect to cranial tibial translation. Both studies noted approximately 10mm of cranial tibial translation after CrCL transection. This change was most pronounced at the beginning of stance and was maintained throughout stance. This translation returned to more normal values during swing suggesting the stifle is not CrCL dependent during this phase. When this change was followed out long term only a 5mm change was noted [3]. However, this change was reflective of persistent cranial tibial translation throughout gait and not normalization of cranial tibial translation. The authors suggest that subsequent meniscal damage may have prevented the stifle from springing back to the normal position during the swing phase producing the persistent cranial tibial translation throughout the gait cycle.

Other studies have kinematically evaluated repair techniques for the CrCL-deficient stifle. In one study kinematic differences were compared at a trot between dogs treated with a TPLO or a cranial closing wedge osteotomy [154]. Patterns of gait were evaluated preoperatively and at 8 and 12 weeks after transection. After TPLO the stifle was held in greater flexion at the end of stance and beginning of swing. This change was only significant at 8 weeks after transection. The opposite was found in dogs treated with CCWO. During the swing phase dogs treated with CCWO were found to carry their stifle in 10 to 20° more extension as compared to the preoperative values. While this study compared pre and post-operative changes, no control using an unrepaired CrCL-deficient stifle was included so conclusions with respect to the specific procedural effects on gait were not determined. A second study kinematically compared all three conditions (intact, CrCL-deficient and repaired), however this study was performed on cadaver limbs in an unloaded state [13]. In this study range of motion, cranial tibial translation and internal rotation were compared in stifles repaired with the medial retinacular imbrication technique (MRIT) or CCWO. Post-procedural TPA ranging from 8 to 15° were achieved using the CCWO. Cadaver limbs were taken from 85 to 150° of extension. After transection no significant difference was noted in cranial tibial translation. No difference in cranial tibial translation was noted after repair using the MRIT. Significant caudal tibial translation, up to approximately 20mm, was noted after repair using the CCWO. This finding is interesting in that the limbs were not loaded and tibial translation is a load generated phenomenon [13]. Range of motion was significantly decreased after repair with the MRIT; range of motion was not affected by the CCWO. Relative external rotation was noted after MRIT and CCWO. With respect to the CCWO the authors suggest external rotation may have been created during fragment fixation.

CHAPTER II: The Effect of Stifle Angle on Stifle Kinematics following TPLO: An *in vitro* Experimental Analysis

A: Objectives

The objectives of this study were to (1) evaluate cranial tibial translation and (2) internal rotation in a TPLO repaired stifle through a loaded range of motion; and (3) to evaluate the effect of the TPLO on hyperextension in a CrCL-deficient stifle. We hypothesized that the TPLO would re-establish normal kinematics to the CrCL-deficient stifle throughout the range of motion.

B: Materials and Methods

1. Specimen Preparation

Twenty-four canine hind limbs were harvested by coxofemoral disarticulation from 12 dogs weighing >20 kg that were euthanized for reasons unrelated to the study. Caudocranial and mediolateral radiographs were taken of each stifle joint to confirm skeletal maturity and absence of orthopedic disease. The mediolateral radiographic view was used to measure the tibial plateau angle (TPA) of each stifle as previously described [98]. The degree of rotation required to achieve a post-procedural TPA of 5° was determined using a previously described method [98]. After radiographic imaging, the limbs were stripped of all musculature proximal to the talocrural joint. Only the gastrocnemius muscle was preserved. The non-muscular soft tissues of the stifle joint (cruciate ligaments, collateral ligaments, menisci and joint capsule) were also carefully preserved. After dissection the limbs were wrapped in saline (0.9% NaCl) soaked towels and stored at -10°C until biomechanical testing.

At the time of testing the limbs were thawed at room temperature for approximately 12 hours. Throughout testing isotonic saline was sprayed on the limbs intermittently to keep the tissues moist. The femoral head and greater trochanter were removed en bloc at the level of the third trochanter. The proximal 1/3 of the femur was then placed eccentrically in a 4-inch length of 2-inch diameter PVC pipe. Polymethymethacrylate (PMMA) was used to secure the limb within the PVC pipe. After potting, a 2.5mm hole was created in a cranial to caudal direction through the widest portion of the patella.

Eighteen-gauge orthopedic wire was passed through the hole and the ends were twisted together creating a distinct wire loop. Electromagnetic motion-tracking sensors with six degrees of freedom (FasTrak, Polhemus; Colchester, Vermont, USA) were placed on the lateral aspects of the distal femur and proximal tibia and secured, each with two ¾-inch wood screws. A double-acting pneumatic cylinder with 38mm diameter bore and 127mm stroke was secured to the cranioproximal aspect of the femur using 2 hose clamps around the PVC pipe and cylinder. Eighteen-gauge orthopedic wire was then passed through a right-angled extension of the piston and then through the loop of wire around the patella before being secured with multiple twists. The cylinder was aligned in a way such that retraction of the piston applied a force parallel to the femur mimicking the action of the quadriceps muscle. Force applied by the cylinder was manually controlled using a custom-built electropneumatic regulator. Limbs were then placed in a custom-designed loading frame, which has previously been used to simulate the stance phase of canine gait [155] (Figure 1). The pes was secured at the level of the metatarsals in a custom designed boot that maintained the pes at an angle of approximately 70° with the horizontal. The combined femur, PVC pipe and cylinder were secured to the proximal plate of the jig with a shoulder bolt, which essentially created a hinge joint to confine motion of the proximal femur to the sagittal plane. The proximal plate slid smoothly (via linear bearings) on 4 precision ground shafts so that its weight simulated the superincumbent body weight. By placing weights on top of the plate the force supported by each limb was adjusted to 20% body weight for each subject.

2. Biomechanical Testing Protocol

With the intact limb mounted in the loading frame, force on the cylinder was adjusted to achieve static equilibrium with the stifle at 120° extension, as determined using a goniometer following the diaphyses of the femur and tibia. The angular orientation (azimuth, elevation, and roll) of the femoral and tibial sensors was zeroed at this position. The pneumatic cylinder force mimicking the quadriceps mechanism was then steadily increased at approximately 10N/second until the stifle angle moved from 120° to maximum extension. The force threshold at which motion occurred was usually 300-400 N. Data was collected using the 6 degree-of-freedom electromagnetic motion-tracking sensors at a rate of 60 points/second for each sensor throughout the range of motion. The limb was then unloaded and the procedure was repeated twice for a total of three trials. The limb was again unloaded and a 2 cm medial arthrotomy was performed. A nerve hook and a 64 mm beaver blade were used to isolate and transect the CrCL. Transection was confirmed by visual inspection and presence of palpable cranial drawer. The limb was then reloaded and taken through another 3 trials as before. Data was again recorded throughout the range of motion for the CrCL-deficient stifle. The limb was then removed from the loading frame, the motion-tracking sensors were removed and a TPLO was performed as previously described [9,98]. Radial osteotomies were rotated as previously calculated to achieve a TPA of approximately 5°. Osteotomies were secured with 3.5 mm TPLO plates (Securos Veterinary Orthopedics; Fiskdale, Massachusetts, USA) using 3 self-tapping cancellous screws proximally and 3 self-tapping cortical screws distally. Absence of cranial tibial thrust was confirmed via palpation. The motion-tracking sensors were again secured in the same orientation as before using the previously drilled holes in the femur and tibia, the limb was replaced in the loading frame and the limb was tested as before.

3. Data Analysis (to standardize tibial thrust and internal rotation)

The source for the eletromagnetic motion tracking sensors was placed on the left side of the jig at approximately the level of the distal femur. This resulted in a coordinate system in which the positive x-axis extended caudally, the positive y-axis extended proximally, and the positive z-axis extended to the right. Values for x, y and z were collected during testing along with values for roll, elevation, and azimuth corresponding to rotation about the x, y, and z-axes, respectively. The motions of interest were CTT and internal rotation throughout the range of motion as well as the maximum angle of extension. Joint angle at any given point was calculated by adding the absolute values of the femoral and tibial azimuth values. As these values were zeroed at 120° of extension during testing 120° was added to each of these sums. CTT was calculated from the difference in x values of the tibial and femoral sensors. Internal rotation was quantified from the difference in femoral and tibial elevation values. As the joint angle and internal rotation orientation values were zeroed at the start of testing only relative change throughout the range of motion could be evaluated. This deficit in data collection was identified late in testing. However, for 5 limbs absolute values within the coordinate system were recorded prior to zeroing. The median values for each condition for each limb were statistically compared to determine if there was a significant difference between the initial internal rotation coordinates for each condition.

Graphs were then created for each trial for each condition for joint angle versus CTT and joint angle versus internal rotation. Joint angles from 120 – 145°, corresponding to the weight bearing range of motion during a walk [78], were evaluated. The data was then curve-fit with a first order polynomial. If the R^2 value for the trend line was <0.98 the data was fit to a second order polynomial. If the R^2 value was still <0.98 a third order polynomial was created. No further improvement in R^2 values was attempted with higher order polynomials. The majority of the data required second order polynomials. These trend lines were then used to calculate the CTT and internal rotation values from 120 to 145°. During graphic evaluation it was noted that a single trial would occasionally (<8%) be inconsistent with the other two. To minimize the effect of any outlying trials median values for each of the three trials were used for data analysis.

4. Statistical Analysis

A mixed model ANOVA (followed by Tukey's procedure for multiple comparisons) was used to evaluate differences in CTT and internal rotation among the three different conditions (intact, transected, and TPLO repaired) at each angle throughout the loaded range of motion. Treatment (intact, transected, and TPLO repaired) and angle (120° – 145°) were fixed effects. Dog specimen and side within the dog (right or left) were specified as random effects (ie, measurements within dog specimen and side within dog were not independent). Furthermore, a scatter plot (and bar graph) for mean relative CTT vs angle for each of the conditions (CrCL-deficient or TPLO repaired) showed a curvilinear relationship. Subsequently, this relationship was modeled using polynomial regression. Statistical significance was defined as a one sided p-value less than 0.05 for the ANOVA and a two sided p-value for polynomial regression. Maximal values for extension were evaluated using a mixed model ANOVA. The 5 limbs with initial absolute orientation values for internal rotation were also evaluated using a mixed model ANOVA. A statistical analysis software package (SAS/STAT software, Version 9.2 of the SAS System for Windows, SAS Institute Inc.; Cary, North Carolina, USA) was used to perform all statistical analyses

C: Results

Median body weight of the cadavers was 23.75 kg (range 20-40 kg). Median TPA was 22° (range 16-26°). To achieve a post-operative TPA of 5°, the median displacement cord along the osteotomy line was 7 mm (range 4.5 to 8.25 mm).

1. Cranial Tibial Translation

After CrCL transection, CTT increased significantly ($p < 0.001$) throughout the entire loaded range of motion with a relative median value of 5.92mm (range 4.98 – 6.08mm). Values tended to decrease as the angle of extension increased (Figure 2). Relative CTT also occurred in the TPLO repaired stifle with a median of 1.98mm (range 1.26 – 3.36mm). Values tended to increase as the angle of extension increased. These TPLO CTT values trended towards values of the intact stifle (Figure 2). There was no significant difference ($p > 0.05$) in relative CTT values between the intact and TPLO repaired stifle from 120° to 130°. This difference was significant ($p < 0.03$) from 131° to 145°. Despite failure of the TPLO to normalize CTT throughout the entire range of motion, TPLO values were significantly lower (<0.04) than CrCL-deficient values at all angles excluding 145° ($p = 0.05$).

For the CrCL-deficient stifle, the curvilinear relationship between mean relative CTT and angle was estimated as: Mean CTT = $-59.6 + 1.02$ Angle – 0.004 Angle^2. Both the linear ($p<0.0001$) and quadratic ($p<0.000$) angle terms were statistically significant with an R-squared of 0.9993. For the TPLO condition, the curvilinear relationship between mean relative CTT and angle was estimated as: Mean CTT = $49.3 + -0.8$ Angle + 0.003 Angle^2. Both the linear ($p<0.0001$) and quadratic ($p<0.000$) angle terms were statistically significant with an R-squared of 0.9995.

2. Internal Rotation

Minimal relative tibial rotation occurred throughout the loaded range of motion in both the CrCL-deficient and TPLO stabilized stifle. For both conditions internal rotation tended to increase with increasing angles of extension. However, these values were minimal (Figure 3). The median value for the CrCL-deficient stifle was 0.61° (range -0.19° – 1.34°) throughout the range of motion. The median value for the TPLO stabilized stifle was 0.05° (range -0.10° – 0.54°). All three conditions were not statistically different ($p > 0.05$) from 120° to 137°. The intact and CrCL-deficient internal rotation values were significantly different ($p < 0.05$) from 138° to 145°.

Evaluation of the absolute internal rotation orientation values at 120° of extension did not identify a statistical difference between any of three treatment groups. This finding suggests that the lack of statistical difference at 120° seen in the relative values is a true finding and significant data was not lost with zeroing at the beginning of each trial.

3. Hyperextension

During testing it visually appeared as though there was a greater degree of extension after repair with the TPLO. This difference was not quantified using a goniometer. The median final joint angles for the intact, CrCL-deficient and TPLO repaired stifles were 161.67°(range 147.29 – 179.17°), 163.67° (range 151.06° – 190.45°) and 162.66° (range 150.26° – 196.74°), respectively. When values were compared between conditions no significant differences were identified for maximum extension angle ($p > 0.05$).

D: Discussion

As described by Arnoczky et al the three main functions of the CrCL include prevention of CTT, excessive internal rotation and hyperextension [1]. It has also been shown that these functions are most essential during the stance phase of gait, the weight-bearing portion of gait

[2]. An *in vitro* model capable of mimicking a loaded range of motion would allow for direct comparison of the 3-dimensional changes occurring in the CrCL-deficient stifle with various surgical techniques. Alterations from the normal state could help to identify influential components in the progression of osteoarthritis and potentially provide insight into modifications that may better re-establish normal biomechanics in the joint. To the author's knowledge, no study has assessed the effect of a CrCL reconstruction technique on the 3-dimensional kinematics of the stifle throughout a range of motion in a loaded state. In this study a loaded weight-bearing model was used to evaluate these 3-dimensional changes and the associated effect of repair using the TPLO. In our model, a force mimicking the quadriceps was applied along the axis of the femur which allowed the stifle to move from an angle of 120° to maximal extension while loaded to 20% body weight. While this model greatly simplifies the mechanical forces acting upon the stifle, it provided a repeatable method for comparing three different conditions with the same specimen serving as its own control. Additionally, use of the electromagnetic sensors for evaluation of the 3-dimensional changes provided an accurate and reliable method for comparing parameters between conditions.

1. Cranial Tibial Translation

In this study CTT was significantly increased in the CrCL-deficient stifle throughout the entire range of motion as compared to the intact stifle. This load generated CTT has been reported in other studies at a single standing angle [10-12]. Warzee et al reported a relative increase of 18.9mm +/- 3.4mm at a standing angle of 135° [10]. Reif et al found a relative mean increase of 14 +/- 2.3mm at a standing angle of 120° [11]. Kim et al noted a relative change of 15.3mm at 135° of extension [12]. While our median values at 120° and 135° were approximately a third of these values (5.8mm and 5.9mm, respectively) the general concept of generation of cranial tibial translation with load was demonstrated. These smaller values were likely due to more proximal and distal positioning of the markers on the femur and tibia, respectively. These other studies evaluated cranial tibial translation through quantification of the relative change in position of the insertion of the CrCL.

In our model the more distal location of the sensor evaluated a position on the tibia that anatomically translates to a lesser degree than the articular surface of the tibia

Few studies in the veterinary literature have evaluated the change in CTT throughout a range of motion in a CrCL-deficient stifle. In a non-weight bearing biomechanical model of CrCL rupture CTT was only noted from 135° to 150° of extension with a maximum of approximately 1.5 mm at 150° [13]. These values did not differ significantly from the intact stifle. As CTT is generated by weight bearing the lack of significant translation is understandable. Korvick et al used a spatial linkage system to evaluate the 3-dimensional changes in live dogs throughout the entire gait cycle and found that transection of the CrCL resulted in a progressive change from approximately 4 to 10 mm as the stifle was flexed during the stance phase [2]. Tashman et al reported similar results with approximately 5mm of change with flexion [3]. This general trend towards increasing CTT values with increasing angles of flexion was noted in our results (Figure 2). As this trend has also been demonstrated in the anatomical cruciate study by Arnoczky et al [156], in both the intact and cruciate deficient stifle, it is likely that CTT is controlled by other anatomical components of the stifle joint at higher angles of extension.

While the results of this study are in agreement with previous studies comparing the intact and CrCL-deficient stifle, the results of the TPLO repaired stifles are more conflicting. In the static weight-bearing models CTT is converted into caudal tibial translation after TPLO. The magnitude of this caudal translation increases with decreasing tibial plateau angles as shown in the study by Warzee et al [10]. In that study caudal tibial translation increased by approximately 3mm as the tibial plateau was rotated from 6.5° to 0°. In our model CTT was present throughout the entire loaded range of motion despite rotation to a TPA of approximately 5°. This is likely due to the active quadriceps force applied to extend the knee. It is possible that this generated enough force to resist the caudal tibial translation and potentially promote CTT. Alternatively, our results may in fact represent the *in vivo* condition. In an extensively researched 3-dimensional mathematical model of the stifle joint it was determined that strain within the CrCL is minimal in a TPLO repaired stifle rotated to 0° [127]. However, in a TPLO repaired stifle rotated to 5° significant strain was still present in the CrCL throughout the entire stance phase of gait. Theoretically this strain represents a persistent cranially directed shear force that could contribute to the continued presence of CTT in a TPLO repaired stifle rotated to 5°. While results of our study support the findings of this mathematical model, it is recognized that neither a mathematical nor a biomechanical model fully addresses all components of the *in vivo* condition. The superiority of the mathematical model over *in vitro* studies is not known.

Despite the continued presence of detectable CTT the values in the TPLO repaired stifle were not significantly different from the intact stifle from 120° to 130° of extension indicating that the TPLO effectively normalizes the CTT resulting from CrCL deficiency. From 131° to 145° there was a significant difference from the intact stifle suggesting persistently altered biomechanics. If the data is reviewed using $p < 0.01$ there was no significant difference between the intact and TPLO repaired stifles from 120° to 138° indicating the TPLO may restore normal biomechanics through a larger range of motion as modeled in this study. Similarly, although the TPLO did not normalize CTT throughout the entire range of motion it trended towards intact values at all angles excluding 145°. From 120° to 144° the TPLO values were significantly less than the CrCL-deficient values as demonstrated in Figure 2.

This trend towards diminished effectiveness in attenuation of CTT at larger angles of extension is not a surprising consequence of the TPLO. In a normal stifle the point of contact between the femoral and tibial articular surfaces through which load is transmitted is cranial to the functional axis of the tibia [157]. As a result, loading of the joint results in a force vector that contains a ventrally directed compressive force as well as a horizontally directed force, termed tibial thrust. Due to the caudal slope of the tibial plateau with respect to the functional axis of the tibia, tibial thrust promotes CTT during loading [157]. In a normal stifle the CrCL counteracts this force vector. In the absence of this ligament CTT occurs. The principle behind the TPLO is based on rotation of the tibial plateau such that the point of contact between the articular surfaces of the femur and tibia lies along the functional axis of the tibia [84,157]. As a result, loading of the joint results in a purely compressive force vector eliminating the need for a CrCL. However, alignment of joint load along the functional axis of the femur will only occur for a small range of motion. This point of contact between the articular surfaces is dynamic throughout motion. As the stifle extends this point moves cranially; as the stifle flexes this point moves caudally [84]. Thus although joint load is converted to a single compressive vector with the TPLO at a specific angle, with movement away from this point joint load is again composed of two vectors. As the stifle is flexed in relation to this neutral point, contact between the articular surfaces moves caudally and the tibial slope becomes positive in relation to the functional axis of the tibia. As a result, the horizontally directed force vector points in a caudal direction. In a stifle with an intact caudal cruciate ligament this force is neutralized. Thus at angles less than 135° the stifle is still functionally stable with respect to cranial-caudal motion. However, as the stifle extends from the neutral point, contact between the femur and tibia moves cranially and the tibial slope again becomes negative with respect to the functional axis of the tibia. This horizontal force vector is directed cranially resulting in unapposed cranial tibial thrust with subsequent CTT. In this study, if we evaluate tibial translation with respect to 135° as the stifle flexes CTT decreases; as the stifle extends CTT increases. Thus, as demonstrated by the progressive loss of significance with extension in our model the TPLO is more effective at normalizing CTT at angles of greater flexion than extension.

This effectiveness with respect to angle of flexion is of note when *in vivo* cruciate deficient models are considered. These studies have shown that dogs compensate for loss of the CrCL by holding the stifle in a greater degree of flexion [2,3]. By carrying the stifle in a greater degree of flexion the slope of the tibial plateau with respect to the functional axis of the femur is minimized, potentially allowing for less CTT throughout stance and thus providing a greater degree of stability to the knee. In the TPLO repaired stifle the procedure itself places the stifle in a greater degree of flexion with respect to the tibial plateau [98]. If dogs then further compensate for CrCL deficiency via increased flexion the angular excursion during gait is altered. For example, if the normal stifle flexes from 120-145° during the stance phase [78] and compensation of approximately 10° is applied the range of motion now occurs from 110° to 135°. At these lower values the TPLO is better able to compensate for the loss of the CrCL. Thus, with an added compensation in gait the TPLO may effectively restore normal CTT throughout stance.

2. Internal Rotation

In our model minimal relative internal rotation was noted. From 120° – 137° there was no statistical difference between the intact, CrCL-deficient and TPLO repaired stifle. These results are dissimilar to those reported in other studies. Passive manipulation has been shown to increase internal rotation by a mean of 11° in extension and 26° in flexion after transection of the CrCL [1]. Similarly, most weight bearing models report internal rotation with rupture of the CrCL. An *in vivo* weight bearing research model identified a significant increase in internal rotation after transection of the CrCL throughout the stance phase of gait [2]. The degree of internal rotation was more pronounced at greater flexion angles. Static weight bearing studies examining the TPLO also demonstrate the presence of increased internal rotation after CrCL transection with normalization after TPLO. One study found that the unrepaired CrCL-deficient stifle resulted in $23.2^o \pm 6.8^o$ of internal rotation when the stifle was loaded at a weight-bearing angle [10]. Stabilization with the TPLO rotated to a TPA of 0° subsequently reduced internal rotation to $9.5^o \pm 6.3^o$. This difference was significant. In a more recent *in vitro* study internal rotation in the CrCL-deficient stifle was found to have a relative mean rotational change of 14.3° [12]. This value was attenuated to a relative difference of -1.2° in the TPLO treated stifle rotated to 6°, indicating relative external rotation. There was no significant difference between the intact and TPLO repaired stifle with respect to internal rotation suggesting that the TPLO is capable of restoring this aspect of normal kinematics.

While our model also demonstrated a lack of significant difference between the intact and TPLO repaired stifle, the failure to demonstrate a difference between the intact and CrCL-deficient stifles as demonstrated in these other studies brings into question the validity of this model for evaluation of internal rotation. A plausible cause for this result lies in the application of the quadriceps force. If the force mimicking the quadriceps muscle was not directly in line with the axis of the femur, but instead more medial or lateral, application of this force would result in an active force in the tibial rotational plane. This force could then either promote or resist internal rotation. As this direction of force was not altered within a limb each condition should have been affected similarly. However, in an intact stifle the normal cruciate likely over rode this force. In the unrestrained CrCL-deficient conditions this force could have been more influential. Thus if by chance similar numbers were positioned to the right or left, the higher or lower values for internal rotation would have averaged to a seemingly more unaffected value, resulting in significant type II error with this data set. Alternatively, some may argue that minimum internal rotation was recorded in all conditions due to constraint of the pes. In our model coxofemoral motion was limited to flexion-extension and the pes was held in contact with the loading surface. Application of the quadriceps force resulted in extension of the stifle and the hock. Motion of the tibia was defined proximally by the passive restraints of the stifle and distally by the ligamentous constrains of the hock. As, to the author's knowledge, no studies have been performed that assess the effect of hock angle and associated ligamentous constraints on proximal tibia motion this potential effect on internal rotation is purely speculative.

However, results of our study may in fact support the clinical picture. In an *in vivo* kinematic study no difference in degree of internal rotation was noted between the intact and CrCL-deficient stifles [12]. They suggest that gait in itself may not generate sufficient torques during weight bearing to generate internal rotation, thus limiting the internal-external rotational control of the CrCL to the swing phase. They also suggest the degree of rotation may be dictated more by the articular geometry and soft tissue components of the joint than by the CrCL during the stance phase. Our results support this theory as internal rotation was not different between the intact and CrCL-deficient stifle suggesting that transection of the ligament does not result in significantly altered rotation. Additionally our results suggest the TPLO does not alter the surface of the tibial plateau such that abnormal rotations are created. Although this justification is consistent with our results, with many more studies supporting the role of the CrCL in limiting internal rotation during the stance phase, failure to re-create normal biomechanical conditions is possibly a more likely explanation for our results.

In addition to error attributable to model design, errors in data collection may also have contributed to the unexpected internal rotation results. During testing the rotational values about the x, y and z axes were zeroed so at to establish that position as 120° of limb extension. As a result the initial internal-external rotation were zeroed. While the behavior of each condition could be compared as the limb was extended, any initial absolute difference in internal rotation was lost. To compensate for this loss, the initial non-zeroed coordinates were recorded for 5 limbs. These initial values were statistically evaluated and no difference was identified indicating that the observed lack of difference at the initial value of 120° was in fact the true case. However, with only a small number of samples it is possible that a significant difference remained undetected. Collection of absolute coordinate values prior to zeroing for all limbs would have improved this result and made comparisons more valid.

3. Hyperextension

The role of the CrCL as the primary check against hyperextension of the stifle is a generally accepted concept in veterinary medicine. This function is based on the anatomic study performed by Arnoczky et al that found the mean angle of extension with passive manipulation increased from 148° to 160° after transection of the CrCL [1]. Smith et al similarly reported a range of 140-150° in the intact stifle with a 10° increase after CrCL transection [158]. Few studies since have included assessment of hyperextension in the CrCL-deficient stifle. This is likely due to the absence of hyperextension and related injuries during normal gait. However, as we aimed to evaluate the three basic functions of the CrCL results for hyperextension were included. In our study there was no significant difference in maximal extension angle between the intact, CrCL-deficient and TPLO repaired stifle. The lack of difference between the intact stifle and the CrCL-deficient stifle suggests that in a loaded model absence of the CrCL does not result in hyperextension. Perhaps in a loaded state hyperextension is actually prevented by other anatomical structures and is not reliant of the CrCL. Further studies would be necessary to investigate this possibility.

Hyperextension was also evaluated as subjectively it appeared as though the TPLO repaired stifles achieved a higher degree of extension than the intact and CrCL-deficient stifles. In theory as the TPLO is placing the articular surface at a greater angle of flexion one could rationalize that the TPLO would allow a greater degree of extension. However, statistical analysis did not support this observation. As a goniometer was used to establish the initial angle of 120° perhaps evaluation using a goniometer, instead of the calculated angular position of the sensors, to measure maximal extension would have resulted in more significant, although likely less accurate, differences.

4. Methodology

The major limitations in this study lie in its *in vitro* nature. The dynamic forces acting on the stifle were reduced to the single effect of the quadriceps muscle. While this allowed for simplification in model design it did not recreate the *in vivo* condition. The effects of other important musculature as well as the neurovascular components contributing to the stability of the stifle were not assessed. *In vivo* these factors may alter the kinematics of the stifle thus making results from our simplified model an inaccurate assessment of the true condition. Additionally the limbs in this study were harvested from normal dogs. In the clinical situation a dog presenting with a CrCL rupture inevitably has significant degenerative changes within the stifle as well as various degrees of peri-articular scarring and muscle atrophy. These changes can affect the kinematics of the stifle. This was demonstrated in a study that evaluated the long-term effect of the transection of the CrCL. With time it was found that CTT became progressively more abnormal as degenerative changes progressed in the stifle [158]. Our *in vitro* study did not account for this variable.

Another limitation in this study was the assumption that abnormal 3-dimensional kinematics suggests abnormal loading. Positional abnormalities between two bones are generally thought to indicate abnormal loading, which, in turn, contributes to progressive osteoarthritis [5]. Re-establishment of normal positioning would theoretically indicate a return to normal loading and as such would be a valuable technique for comparing two conditions. However, in a recent study it was shown that although the TPLO reestablished normal relative positioning of the tibia and femur, load distribution was still significantly altered [12]. This would indicate that although our data suggests the TPLO is capable of minimizing abnormal tibial positioning this can not be extrapolated to indicate a trend towards normalization of loading. Further studies are necessary to better evaluate the effect of the TPLO on load distribution throughout weight bearing.

Lastly, while we evaluated a continuous range of motion inclusive of the weight-bearing angles we did not evaluate them in the normal gait sequence. Our model more accurately represents the terminal portion of a sit to stand motion. Load during gait progressively increases then decreases as the limb accepts and transfers weight, respectively [12,159]. Joint stability has been shown to be positively affected by joint load [160]. As the limb moves through the stance phase the magnitude of joint load becomes more influential on the kinematics of the stifle. As this was an initial attempt to model weight bearing this load was kept constant to simplify our model. Thus, extrapolation of our data to the stance phase of gait must be interpreted with caution due to the failure of our model to account for this variable. Future *in vitro* models should consider this variable to more accurately represent the *in vivo* stance phase.

5. Future Directions

One significant concern in this study is the lack of significance in internal rotation values for all conditions tested. Although many different factors could be influential one factor could easily be evaluated in a future study. Our study could be repeated with an unconstrained pes to evaluate this potential effect of the present construction on internal rotation. This would allow for objective evaluation of this potentially influential factor. Additionally using only right or left hind limbs would help to eliminate any effect of limb positioning within the coordinate systems that may have averaged significant rotational values.

Reevaluating our model without the presence of the gastrocnemius would also be interesting. In static weight bearing models a gastrocneumius component is necessary to maintain the limb in a loaded state. However in our model this muscle was not necessary as we applied an active force to the quadriceps and maintained pes support. Although leaving this muscle intact likely did not affect our results, as the active quadriceps force is known to contribute to CTT in a CrCL-deficient stifle, leaving this muscle intact may have altered our results.

Accounting for the limitations of this study future studies could evaluate a multitude of factors associated with CrCL disease. The kinematic characteristics of other commonly performed repair techniques for the CrCL-deficient stifle, such as the TTA and lateral fabellar suture, could be evaluated using our model. Additionally the effects of the medial menicus on stifle stability through a range of motion could be evaluated. For example, limbs with intact CrCLs could be evaluated with an unaltered medial meniscus, after a caudal mensical release and after caudal pole menisectomy. These commonly performed alterations to the medial meniscus could also be tested in combination with various stabilization methods.

To objectively evaluate the clinical effect of the TPLO, or other techniques, on stifle kinematics, *in vivo* studies, such as that performed by Korvick et al [2] using a spatial linkage system, would need to be conducted to assess the effect of the repair technique on stifle kinematics. Studies using normal dogs as well as dogs with CrCL rupture would be most ideal.

E. Conclusion

Results of this study suggest that the TPLO does not fully restore normal kinematics throughout a loaded range of motion. However, TPLO tends to normalize CTT and does not alter internal rotation during the weight bearing range of motion. Additionally the TPLO does not result in more pronounced hyperextension as compared to the intact and CrCL-deficient stifles. Further studies accounting for the potentially influential aspects in our design may help to better classify the effects of the TPLO on normal stifle kinematics under load. A valid weight-bearing model would be paramount in the evaluation and objective comparison of the multitude of techniques employed for repair of the CrCL-deficient stifle. Understanding of these effects on kinematics may help elucidate applicable changes in technique that may help to optimize surgical outcome.

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APPENDIX: Figures

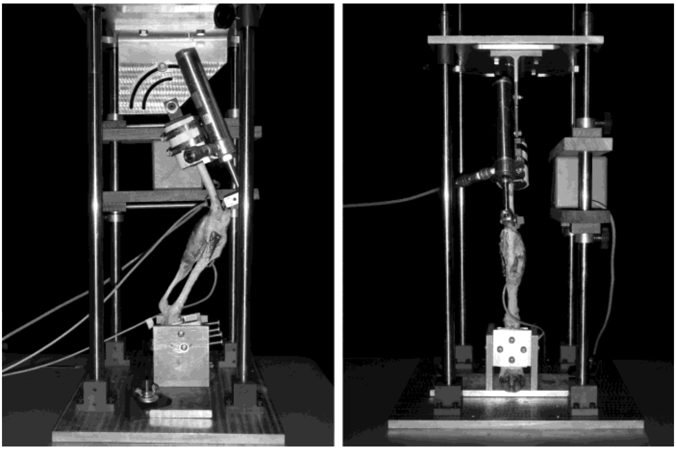

Figure 1. Custom designed loading frame. A TPLO repaired stifle is positioned in the frame. Electromagnetic motion-tracking sensors are secured on the lateral aspect of the distal femur and proximal tibia. The source for the sensors is positioned to the left of the limb at the level of the femur. In the cranial-caudal view the electopneumatic regulator is connected to the pneumatic cylinder on the right side.

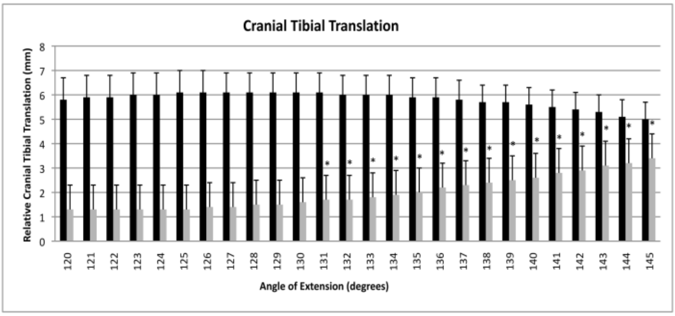

Figure 2. Comparison of cranial tibial translation through a range of joint extension. The x-axis represents the intact condition. The black bars respresent the cranial cruciate ligament (CrCL) deficient condition. The grey bars represent the TPLO repaired condition. Error bars show the positive standard error of the mean for each condition at each angle. There was a significant difference between the intact and CrCL-deficient condition at all angles evaluation. There was a significant difference between the intact and TPLO repaired conditions from 131-145° (*).

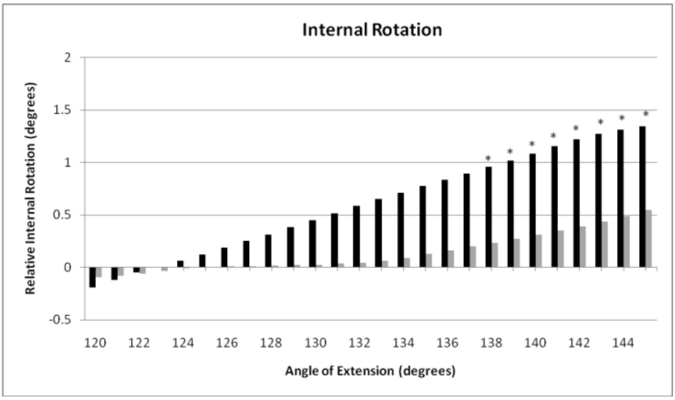

Figure 3: Comparison of interal rotation through a range of joint extension. The x-axis represents the intact condition. The black bars respresent the cranial cruciate ligament (CrCL) deficient condition. The grey bars represent the TPLO repaired condition. There was a significant difference between the intact and CrCL-deficient condition from 138-145° (*).