Biomechanics of the Canine Thoracolumbar Spine in Lateral Bending

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(ABSTRACT)

Pathologic processes and surgical manipulations of the spinal column may result in alterations of the biomechanical properties of the spine through increases or decreases in the range of motion or stability of the spine. A decrease in range of motion between two adjacent vertebrae subsequent to arthrodesis or ankylosis appears, clinically, to be well tolerated without significant alterations to the functions of the spine; however, a decrease in spinal column stability as a result of pathologic changes or surgical alterations can result in catastrophic spinal cord injury.

In order to determine the effect of various surgical procedures and trauma on the spinal column, *in vitro* biomechanical studies may be employed using a servohydraulic testing apparatus and cadaver vertebral motion units. The T_{13}-L_1 vertebral motion units of 48 mix breed dogs were dissected free of surrounding musculature and prepared for biomechanical testing by mounting with cross pins and

polymethylmethacrylate. Specimens were surgically altered by facetectomy, lateral fenestration, diskectomy, and combinations of these procedures. Specimens were subjected to lateral bending at a rate of 2.5 cm per minute to failure in a swing arm bending jig designed to simulate four point bending. The slopes of bending moment vs. angular displacement curves were compared and significance determined by the method of least squares.

A statistical difference ($p < 0.05$) was found between the stiffness of all diskectomy groups when compared to any other group. Unilateral and bilateral facetectomies, and fenestration induced a non-significant decrease in stiffness in comparison to control specimens. This data may be combined with that of previous testing of the canine thoracolumbar spine in flexion-extension and rotation to determine the clinical effects of surgical manipulations and trauma on spinal stability.[1, 2] These results suggest that fenestrations and facetectomies do not appear to increase the risk of injury to the canine thoracolumbar spinal cord during lateral bending in the *in vitro* model; however, thoracolumbar spinal fractures involving the vertebral body as represented by the diskectomy *in vitro* model may significantly destabilize the spine in lateral bending.

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Members of the Schulz family

Eleanor Russell

My clients and patients

My many mentors

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I. LITERATURE REVIEW

The first recorded investigations of spinal stability were by Leonardo Da Vinci who depicted the cervical spine as a ship's mast and the paraspinal muscles as guy ropes providing stability.[3] When stability decreases markedly, the spine can no longer resist forces without significant deformation. Thus, instability may be described as the ability of a small force to cause large or catastrophic vertebral body displacement.[4] White and Panjabi define spinal instability as "the loss of the ability of the spine under physiologic loads to maintain its pattern of displacement so that there is no initial or additional neurologic deficit, no major deformity, and no incapacitating pain."[5] Spinal biomechanics, however, represent a continuum, and therefore, identification of a specific point of instability becomes difficult if not impossible.

Indications for spinal surgery in the dog include treatment of intervertebral disk disease, spinal fractures and luxations, diskospondylitis, and spinal neoplasia. Many surgical techniques that result in anatomic alteration of the spinal column have been described for treatment of spinal disease in the dog. In order to accurately assess the effect of these surgical alterations on spinal strength and stability, a data base of vertebral motion unit stability in normal and surgically altered spines is necessary. There is no objective data relative to canine spinal column strength and stability in lateral bending following surgical manipulation.

Indications for and consequences of spinal surgery should be based partially on a knowledge of spinal biomechanics. The effects of disease processes and surgical manipulations on the stability and function of the spine are based on an evaluation of specific biomechanical parameters of normal and altered vertebral motion units. A vertebral motion unit is defined as two adjacent vertebral bodies and the ligaments, joint capsules and intervertebral disk that connects them. The influence of paraspinal muscles on the unit is disregarded in this model due to the difficulty and inaccuracy of evaluating the biomechanics of muscles in a postmortem specimen. Biomechanical parameters of interest include range of motion, degrees of freedom, coupling, load and displacement.

The effects of surgical procedures such as laminectomies on spinal stability may be determined by comparing the range of motion and load vs. displacement curve of a normal vertebral motion unit to those of a vertebral motion unit modified by a laminectomy. Similarly, the effects of traumatic fractures on spinal stability may be estimated by comparing the biomechanical parameters of a normal vertebral motion unit to those of a vertebral motion unit with simulated injuries.

Knowledge of the biomechanical effects of spinal surgery may aid the surgeon in predicting which surgical procedures can be performed without a marked increase of spinal instability that may result in catastrophic failure of the spinal column. This knowledge may also assist the surgeon in predicting which spinal injuries result in instability that could permit further spinal cord damage and therefore necessitate surgical stabilization.

A. Surgical approaches to the canine spine

Surgical procedures which disrupt or alter the normal anatomy of the spine may alter its biomechanical and protective functions. Laminectomies and intervertebral disk fenestrations performed in the thoracolumbar area are surgical procedures that alter normal spinal anatomy and significantly increase spinal rotational instability.[6-22] These techniques are most commonly employed for treatment of intervertebral disk disease and anatomically are centered over an intervertebral disk space.

Historically, several techniques of laminectomy have been described, each of which alters spinal anatomy in a different fashion. An osteotomy of the articular facets was described in 1951 as a means of thoracolumbar spinal cord decompression.[15] A dorsal laminectomy for treatment of thoracolumbar disk protrusion was subsequently described by Green in 1951[23] and again by Funkquist in 1962. By this technique a dorsal muscle splitting approach is made to the thoracolumbar spine. An ostectomy of the laminae including the articular facets is performed to a level approximating the central canal of the spinal cord (Funkquist method A).[20] This procedure was subsequently modified to retain the cranial articular processes, a portion of the caudal articular processes, and the outer cortex of the pedicles (Funkquist method B). The modification was designed to protect the spinal cord from compression by fibrous connective tissue during healing.[20]

Trotter et. al. described a modified deep dorsal laminectomy with complete removal of the laminae to the level of the dorsal longitudinal ligament and suggested that formation of a laminectomy membrane is not solely dependent on fibrous tissue compressing the exposed spinal cord. The authors suggested that clinically significant spinal cord compression caused by fibrous tissue following laminectomy is partially dependent upon biomechanical alterations subsequent to the surgical procedure.[20] The Funkquist method B was compared to single and multiple adjacent deep dorsal laminectomies (modified type B) by Horne et. al. Neurologic deficits that develop after deep laminectomies over two or more lumbar vertebral bodies were demonstrated histologically to be due to bridging fibrous connective tissue causing spinal cord compression and demyelination. No evidence of spinal instability was noted.[13, 19] Trotter also described a modified dorsal laminectomy whereby the vertebral arches and caudal articular facets are removed but the cranial articular facet remains intact. This procedure was designed to improve spinal cord exposure of the Funkquist method B while preserving the lamina thus offering more protection to the spinal cord.[24]

Redding's initial description of a laminectomy in a dog [15] has since been called a hemilaminectomy, a term which generally implies access to the spinal canal through the right or left lateral lamina. This approach attempts to leave the arches and spinous process intact but may or may not include the articular facets depending on the modifications of the technique. [6, 7, 11, 16, 17, 25] A potential advantage of this technique over dorsal laminectomy is improved exposure of the ventral spinal cord canal while maintaining vertebral column stability.[10] The hemilaminectomy as described by Gage and

Hoerlein [10] involves removal of the dorsolateral lamina from the lateral aspect of the vertebral body to the base of the spinous process. Both cranial and caudal articular processes are also removed over the intervertebral disk space. The selection of dorsal laminectomy or hemilaminectomy in the clinical case is based on the type and anatomical location of the spinal cord lesion.

Intervertebral disk fenestration in the dog was originally described by Olsson in 1951. [26] Fenestration removes the nucleus pulposus of the intervertebral disk in situ by perforation of the outer annulus fibrosus and curettage of the inner nucleus pulposus.[27] The intervertebral disk space may be accessed from a ventral, lateral, or dorsolateral approach and fenestration may be combined with a decompressive procedure. [8, 14, 17, 22] Intervertebral disk fenestration has been advocated for prophylaxis as well as therapy for intervertebral disk extrusion.[17] The clinical efficacy of this technique remains in question. [28] While substantial amounts of nucleus pulposus may be removed, especially when using a high speed air drill[29], other studies have demonstrated that the nucleus pulposus re-forms over time.[30-32]

The efficacy of fenestration as therapy for clinical neurologic deficits caused by intervertebral disk extrusion has not been demonstrated. A proposed mechanism for the alleviation of clinical signs described by Olsson is stabilization of the remaining nucleus pulposus within the disk by induction of inflammation and elimination of further disk prolapse.[8, 21, 33, 34] The effect of fenestration on spinal stability is uncertain.

B. Traumatic spinal injury

Surgical intervention may also be indicated in cases of canine spinal trauma. Luxations and fractures of the canine spinal column may cause severe neurologic abnormalities and extreme pain.[35-37] A large number of dogs with traumatic spinal cord injuries are euthanized because of severity of injury.[35, 38] The major cause of traumatic spinal injury in the dog is motor vehicle accidents while other causes of spinal trauma include dog fights, falling from heights, and gunshot injuries.[35, 38-40] In a retrospective study of 600 motor vehicle accidents involving dogs, 52 suffered vertebral fractures.[41]

Spinal fractures most commonly involve the lumbar region followed by the thoracic and cervical regions. [35, 40, 42] The reason for this distribution is unknown; however the low incidence of reported cervical fractures may be because many traumatic injuries to the cranium and cervical spine are rapidly fatal.[41] The decreased incidence of thoracic vertebral fractures relative to lumbar fractures may be related to the protection afforded to the thoracic vertebrae by the rib cage. Several authors have suggested that there is an increased incidence of fractures at the junctions of relatively mobile and immobile segments of the spine.[27, 39, 40] These sites include the lumbosacral, thoracolumbar, cervico-thoracic, and the cervicocranium junctions. In a review of 51 cases of spinal trauma in dogs and cats, McKee demonstrated an increased incidence of fractures in the cranial and caudal lumbar regions; however another review of 67 cases of fractures of the lumbar spine failed to demonstrate a significant difference in the incidence of fractures between different regions of the lumbar spine.[38]

Spinal fractures in dogs have been categorized into specific fracture types and fracture-luxations. Turner demonstrated in 67 cases of lumbar spinal trauma that the incidence of spinal fractures was twice that of fracture-luxations.[38] Feeney demonstrated an increased incidence of fractures over luxations in the lumbar spine but a higher proportion of luxations in the thoracic spine.[42] In a radiographic survey of vertebral trauma, fractures most frequently involved the vertebral body (50%) followed by the transverse processes(14%), lamina (10%), and articular processes(10%).[42]

Vertebral fractures have been further classified by traumatic force and subsequent fracture type.[43] Ventral compression fractures of the body are theorized to be the result of external forces during flexion of the spinal column. Pure spinal axial compression results in comminuted fractures of the vertebral body while rotational forces contribute to fractures of the articular facets and fracture-luxations. Hyperextension injuries from a dorsal impact are theorized to result in fractures of the laminae and facets. These force-fracture relationships are theoretical and have not been biomechanically demonstrated in the dog. Most spinal fractures are probably the result of a combination of directional forces. The directional forces responsible for specific fracture types are better defined in man. For example, the teardrop or vertebral body wedge fracture that occurs in both man and dog has been demonstrated to be a result of both axial loading and flexion.[44]

Trauma to the spinal cord can result from soft tissue abnormalities that may cause spinal instability or spinal cord compression. Clinical determination of these theoretical soft tissue injuries is difficult and their occurrence has been poorly characterized in the veterinary literature. Traumatic injuries to the intervertebral disk have been described.[40] Luxations of vertebral bodies may involve tearing of the annulus fibrosus, longitudinal ligaments and capsules of the facets. Axial forces may result in traumatic intervertebral disk ruptures and subsequent spinal cord compression. McKee reported positive contrast myelographic findings in 12 cases out of 51 dogs and cats with spinal cord trauma. In five cases spinal cord compression was recognized in conjunction with narrowing of an intervertebral disk space. Laminectomies were performed in some of these cases; however, surgical findings were not reported.[40] Feeney reported 14 possible cases of traumatic disk herniation in 121 cases of spinal cord trauma in dogs and cats. Possible herniation was reported based on narrowing of intervertebral disk spaces.[42] It is unknown whether traumatic disk rupture may induce spinal instability in addition to spinal cord injury.

Therapy for spinal fractures and luxations may include surgical decompression and spinal column stabilization or medical therapy alone. Although definitive indications for surgical intervention have not been established, surgical intervention based on the specific forces involved in the initial trauma has been suggested. Hyperflexion fractures have been categorized as generally stable while compression and rotational forces result in greater instability.[45]

Anatomic compartment classification of injury has been adapted from human biomechanics to aid in diagnosis of spinal instability.[46, 47] In the two compartment model the ventral compartment includes the vertebral body, intervertebral disk, dorsal and ventral longitudinal and intertransverse ligaments. The dorsal compartment includes the laminae, articular facets and spinous processes and the supraspinous, interspinous, and interarcuate ligaments.[46] In the three compartment model a central compartment is defined as the dorsal annulus fibrosus, the dorsal vertebral body and the dorsal longitudinal ligament.[47] Anatomic compartment systems classify vertebral column injury as stable or unstable according to the number of compartments that are involved. In addition to gross vertebral instability, other possible indications for surgical intervention may be the degree of vertebral body displacement and spinal cord compression or spinal cord compression secondary to traumatic disk rupture as suggested by survey radiographs or positive contrast myelography. Clinical indications for surgical stabilization include neurologic deterioration, palpable instability, and radiographic findings.

Radiographic determination of spinal instability is difficult due to the inability to accurately discern soft tissue structures and non-displaced fractures and the inadvisability of performing stressed views to detect movement.[42] Radiographic demonstration of spinal canal compromise has also been shown not to correlate well with prognosis except with severe reduction of spinal canal size.[48] The increased use of computed tomography and magnetic resonance imaging to more accurately assess fracture dynamics and spinal cord compression will be of value.

Numerous stabilization techniques for spinal fractures have been described in the dog[46] and surgical and conservative treatment of canine spinal fractures have been compared.[35, 39] Conservative therapy including spinal splinting and complete exercise restriction have been effective in clinical cases suggesting that no classification scheme can accurately predict which cases will require surgical intervention and that each case must be evaluated independently based on radiographic and clinical findings. A thorough knowledge of the anatomy and biomechanical characteristics of the normal spine aids in decision making.

C. Surgical anatomy of the spine

Spinal stability is maintained by the architecture and strength of the vertebral bodies and the interconnecting annulus, joint capsules, and ligaments. Additional stability may be provided by the surrounding musculature and tendons and the posture maintained by the remainder of the body. The normal canine spine is composed of approximately 35 to 50 interdependent bones. These bones are grouped into 7 cervical, 13 thoracic, 7 lumbar, 3 sacral, and a variable number of caudal vertebrae. The 3 sacral vertebrae are fused early in life while the remaining vertebrae articulate by means of facet joints and disks.[49]

Vertebrae vary in shape, size, and subsequent mobility. Normal vertebrae are longitudinally symmetrical and include a body, laminae, pedicles, and numerous processes. The typical body includes a centrally constricted cortico-cancellous core with cranial and caudal cartilaginous end plates for attachment of the intervertebral disks. The pedicles and laminae are paired and extend dorsally from the body forming the vertebral foramen and the spinal canal. At the cranial and caudal extent of the pedicles are notches which, with the adjacent vertebrae, form the intervertebral foramina which provide access to the canal for the nerve roots, vertebral veins and arteries. The processes which include the articular, transverse, accessory, spinous, and mammillary are points of origin and insertion of the paraspinal muscles.

Thoracic vertebrae are shorter than cervical or lumbar vertebrae and have modified transverse processes for articulation of the ribs at the costal fovea. The dorsal and ventral articular processes of the cervical and cranial thoracic vertebrae change progressively to a medial-lateral orientation beginning cranial to the tenth thoracic vertebra. The subsequent sagittal articulation allows for greater dorsal and ventral movement in the caudal thoracic and lumbar vertebrae. The lumbar vertebrae are typically the most massive in the body and exhibit few variations in shape.

Articulations of the vertebral column include the articular facets, the costovertebral joints, and the intervertebral disks. The facet or zygapophyseal joints and costovertebral joints are diarthrodial. Intervertebral disks contribute approximately 20% of the length of the spine[49] and form amphiarthrodial joints with adjacent vertebrae. The disk is a fibrocartilaginous structure divided into an outer annulus and inner nucleus pulposus. The annulus consists of concentric rings of elastic and collagenous fibers which are firmly attached to the adjacent vertebral body by the cartilage endplates of the vertebral bodies.[50] The dorsal portion of the annulus is one half to one third the diameter of the ventral portion. The nucleus is a gelatinous matrix of collagen, non-collagenous protein and glycosaminoglycans. The intervertebral disks are widest in the cervical and lumbar portions of the spine and function to unite the spinal column, to permit limited mobility, and to absorb shock.[49]

The ligaments of the spine include the nuchal, supraspinous, ventral longitudinal, dorsal longitudinal, interspinous, yellow, intertransverse, and intercapital. The paired nuchal ligaments extend from the spinous process of the axis to that of the first thoracic vertebra. The supraspinous ligament originates on the spinous process of the first thoracic vertebra and inserts on the spinous process of each vertebra to the third caudal vertebra. It limits excessive ventral flexion of the spinal column by preventing distraction of the spinous processes.

The ventral longitudinal ligament attaches to the ventral aspect of the body of each vertebra from the axis to the sacrum. The dorsal longitudinal ligament follows a similar pattern on the dorsal aspect of the vertebral bodies. It is larger than the ventral ligament and extends through the sacrum to the caudal vertebrae. The interspinous ligaments are thin cranio-caudal and dorso-cranial bands of elastic and collagenous fibers extending between spines of adjacent vertebrae.[51] In the lumbar region the intertransverse ligaments span between transverse processes of adjacent vertebrae. The yellow ligament extends between the lamina of adjacent vertebrae within the spinal canal. In the thoracic region the intercapital ligament runs between the heads of the ribs of an individual vertebra. The ligament passes over the dorsal annulus of the disk and under the dorsal longitudinal ligament.[49]

D. General spinal biomechanics

The bony architecture and surrounding soft tissues of the normal canine spine enable it to withstand external forces within a certain range of motion and therefore provide protection for the spinal cord. Diseases and surgical techniques which disturb the spine's function do so by increasing its range of motion and lessening its ability to withstand external forces which lead to deformation and failure. Biomechanical properties of the spine have been determined by examination of vertebral motion units (VMU). These units, composed of two adjacent vertebrae and connecting soft tissue structures, form bone-ligament-bone complexes which possess both material and structural properties. When subjected to a force, the material properties are described as strain relative to stress. Stress is the force per unit area and strain is the deformation per unit length. The vertebral motion unit is composed of numerous components with various material properties.

When viewed as a composite, the structural properties of the VMU are of primary concern. The properties examined are then the deformation relative to force or load.[52, 53] The force is expressed in Newtons and the deformation is expressed in distance. Deformation, expressed in distance or degrees, is then plotted relative to load (force) which is expressed in Newtons. These studies are a component of dynamics or kinetics which describe unbalanced forces on a body and the subsequent motion of that body.

Stress-strain and force-deformation curves may be partitioned into regions as defined on a typical stress-strain curve (Fig. 1). As a load is applied to a structure the initial deformation is reversible and is termed elastic. In a perfectly elastic material deformation is proportional to force as described by Hooke's law. The elastic region may contain both a linear and a nonlinear component. The limit of the elastic region is the yield point or elastic limit. Load applied beyond the yield point produces irreversible or plastic deformation resulting in permanent changes in the structure. The limit of the plastic region is the failure point. Load at or beyond this point causes catastrophic failure and additional load causes no further deformation in a uniform model.[52, 53]

Structural stiffness may be evaluated and compared by use of the slope of load-deformation curves (fig. 2). A steeper slope occurs when a structure demonstrates less deformation in response to load. A structure producing a less steep slope is more easily deformed. Comparisons of the slope of load-deformation curves are the basis of this biomechanical study.

In vitro spinal biomechanics have been performed on intact vertebrae, spinal segments, and vertebral motion units.[54] The vertebral motion unit (VMU) has been used frequently in the study of basic spinal biomechanics. The unit is typically defined as two adjacent vertebrae and their soft tissue connections including ligaments, joint capsules, and the intervertebral disk. The effects of paraspinal musculature and the rest of the body are ignored in the VMU model. The VMU represents the simplest dynamic unit of the spine.[54] Although the orientation of the facets within different regions of the spine limits the directional mobility of the VMU[55], the two body system may be considered to be universally jointed and normally demonstrates 6 degrees of absolute freedom and 3 degrees of relative freedom.[53]

Many biomechanical parameters are concerned with the relative motion of these two bodies which is investigated by application of bending loading. This loading differs from axial, biaxial, torsion or combined loading in the direction or vector of force relative to the loaded object.[52] Bending loading is accomplished by performance of, or simulation of, four point bending.[52] In the four point bending model (fig. 3) the ends of the object, represented as a beam, are supported and the object is loaded centrally, equally at two points. Four point bending induces equal bending moments between the two central loading points, whereas three point bending causes variation of bending moments throughout the beam.

In bending studies load is applied eccentrically and force is a function of the product of the load and the distance from the center of rotation (fig 4). The distance is defined as the moment arm and the product is the moment. The force-displacement curve in bending mode then becomes a moment-angular displacement curve. Biomechanical studies using the VMU model are used to compare force-displacement and moment-angular displacement curves of normal, pathologic, and altered spines as determined by applications of force relative to the normal 9 degrees of freedom.

Application and determination of loads in spinal biomechanical testing is most commonly performed using servohydraulic testing systems. These instruments employ high-pressure hydraulic actuators that apply and record precise loads. The determination of force is made by a load cell which converts load into proportional voltage which may then be recorded by a voltmeter incorporated into a chart recorder or computer. The extent of displacement is simultaneously recorded.[56]

There are numerous limitations to the VMU model. Because the VMU is primarily a tool for *in vitro* analysis it ignores the effects of the rest of the body on biomechanics. Of greatest concern is the contribution of the paraspinal musculature to spinal stability; however, in vitro analysis involving the spine-musculature complex is poorly developed.[56] In spite of this fact, *in vitro* spinal biomechanical testing remains a valuable tool due to the difficulty of *in vivo* experimentation, particularly in humans.[56]

Movements of the *in vivo* normal human spine do not generally occur within single planes or degrees of freedom. Many spinal movements are coupled with others such that in response to a load, movement occurs around more than a single degree of freedom. [54] For example, lateral bending in the normal spine does not occur alone but probably necessitates some degree of concurrent axial rotation. [57] Testing VMUs with servohydraulic systems inherently places limitations on movement within the two vertebrae system. Most mounting systems do not allow free movement within the 3 degrees of relative freedom of the VMU. These restrictions, therefore, lessen the model's ability to simulate the spine's response to external load. The isolation of degrees of freedom is referred to as testing in mode and variations of testing modes have been used to explain the differences in results among researchers regarding spinal biomechanical results.[58]

VMUs fail in other ways to simulate *in vivo* spinal biomechanics. Most *in vitro* biomechanical studies investigate elastic properties of the spine by evaluation of load-displacement curves. Variations in the rate of loading may cause significant differences in results. [56] Investigation of elastic properties also may fail to take into account viscoelastic properties. These include the time dependent effects of creep, hysteresis, and relaxation. These effects are the results of mechanical cycling which may occur up to three million times a year in the human spine.[59]

Instrumentation may cause further problems in the application of *in vitro* biomechanics to the *in vivo* spine. Load cells exhibit cross-sensitivity to secondary forces which may alter results. In addition, fixtures necessary to connect the spinal segment to the materials tester may alter bending moments and limit physiologic movement.[56, 60]

Other parameters that may be determined by *in vitro* spinal biomechanics include range of motion, ultimate failure of spinal segments, instantaneous axis of rotation, and the determination of material properties of the individual components of the VMU. The instantaneous axis of rotation (IAR) is a theoretical center of rotation in a specified plane of bending. Particles at this point exhibit zero velocity during the described bending; understanding changes of this point helps to explain changes in stresses in the spine.[54] The *in vitro* IAR is termed the center of rotation and correlation between the IAR and center of rotation may be difficult.[55]

Knowledge of the material properties of components of VMUs enables the development of material and mathematical models. Normal and abnormal spinal biomechanics may then be approximated by physical testing of material models and computer simulations with mathematical models.[55, 61, 62] The biomechanical properties of a structure ultimately depend upon the organization and material properties of its components.

The vertebral motion unit is composed of bone, cartilage, joint capsule, and ligamentous structures, including the intervertebral disk. For both hard and soft connective tissues, these properties depend upon their viscoelastic characteristics and therefore on loading rates.[52]

Bone is an anisotropic material which is strongest in compression, weakest in shear and intermediate in tension. Bone exhibits a high degree of stiffness in comparison to other biologic materials and a small range of plastic deformation. The materials properties of the soft tissues of the vertebral motion unit vary depending on the amount and structure of collagen, elastin and mucopolysaccharide. Most soft tissue structures are strongest in tension with minimal resistance to shear or compression.[52]

E. Comparative spinal biomechanics

Spinal biomechanics have been extensively investigated in man primarily in response to a high incidence of low back disorders among individuals in developed nations.[4] In the United States, spinal diseases are the greatest cause of chronic disability of individuals under 45 years old. A majority of studies in human biomechanics involve the spine's ability to withstand axial forces in association with the erect posture of humans. The difference in posture and subsequent internal and external forces on the spine make the transfer of human biomechanics to the canine spine difficult. Axial compression is the force of greatest concern in humans while the vector of greatest forces has not been determined in the canine. The importance of normal and pathologic curvature of the human spine also confound the transfer of human spinal biomechanics to canines. Information regarding the material properties of the human spine may, however, be more easily assimilated into canine biomechanics.

Although limited research has been performed in the veterinary community to investigate canine spinal biomechanics, the canine spine has been used as a model for human spinal biomechanics. Zimmerman has compared the mechanical properties of canine and human lumbar motion segments.[63] In examinations of the lumbar spine and intervertebral disk it was found that the compressive stiffness of the lumbar intervertebral disk was similar in man and dogs although canine lumbar disks can withstand greater forces in rotation. Zimmerman also reported that the contributions of the ligaments and articular facets to spinal stability in axial compression, torsion, and flexion-extension are similar in the human and canine spine.[63]

Ranges of motion and load displacement curves comprise essential basic knowledge for in vitro spinal biomechanics. The range of motion in flexion, extension, lateral bending, and torsion has been investigated for every vertebral motion unit of the human spine.[4] Investigative techniques have included the use of goniometry, pantography, and radiographic studies.

In man the load displacement curves of the cervical, thoracic, and lumbar spine have been investigated in compression, shear, anterior, posterior, and lateral bending and in axial torsion.[4] Results of investigators have varied due to alterations in testing mode, coupling, and testing apparatus. Within the thoracic spine, physiologic movements are restricted by the rib cage, particularly in flexion where the rib cage diminishes the range of motion by 70 %.[64] This limitation of movement is not diminished with the removal of one or two ribs unless the sternum is disrupted as well. In man the decreased range of motion in flexion of the thoracic spine has been demonstrated *in vitro*; however, biomechanical studies have not demonstrated increased stiffness of the thoracic spine relative to the lumbar spine in any other plane of bending or rotation. This evidence challenges the theory that a change from a relatively mobile to immobile portion of the spine explains the high incidence of thoracolumbar fractures in humans.[65]

Large axial forces of the human spine are borne primarily by the intervertebral disk and the trabecular bone of the vertebrae. Failure in tension most commonly occurs at the hyaline cartilaginous vertebral end plates; disk failure rarely precedes end plate failure with any forces. The cortical bone of the vertebrae may contribute as little as 10 % of the compressive strength of the spine.[66] The overall strength of a vertebral body may be estimated by measurements of thickness of the cortical ring and the star volume of the trabecular bone. Development of the star volume measurement has allowed estimations of decreases of vertebral strength with age and disease.[67] The vertebral strength in compression of the thoracic vertebrae may decrease by as much as 50 % by 40 years of age.[68] Recently, application of star volume has been applied to clinical cases by means of quantitative computed tomography.[69]

The intervertebral disk is the largest avascular structure in the human body and is subjected to large compressive forces in addition to forces in shear, bending, and rotation. Vertebral bodies have a greater ability than intervertebral disks to withstand tension forces, while the reverse is true for compressive forces.[70] Primary failure of the annulus is not recognized in clinical human medicine. The intervertebral disks absorb forces in bending, compression, and to a lesser degree, tension and shearing and their viscoelastic characteristics may be simulated by material models.[71] Intervertebral disks exhibit less stiffness in tension when failure is more likely.[68] The response of the disk to compressive forces varies with hydration status of the nucleus[4], and degenerative changes of the intervertebral disks appear to lead to compensatory changes in the bone of the adjacent vertebrae.[72] The effects of disk degeneration on spinal biomechanics are still uncertain.[4, 62]

The importance of the intervertebral disk for spinal stability is well recognized. Partial diskectomy by removal of the nucleus pulposus has been used for treatment of disk protrusion and many of these surgeries resulted in worsening of clinical signs.[73] Use of *in vitro* VMUs finite element models of vertebral motion units have suggested that these failures may be partially due to subsequent instability and changes in stress distribution.[74-76]

Because of the asymmetry of the annulus fibrosus, the canine intervertebral disk has different properties in different bending modes. The canine lumbar disk is 30 % stiffer in flexion than in lateral bending.[77] Lumbar diskectomy, in combination with facetectomy, causes a significant decrease in rotational stability of the canine spine.[1] Spondylosis and age-related changes in the intervertebral disks appear to alter the biomechanics of the canine spine resulting in decreased stiffness over time. [78] Surgical alterations of the canine intervertebral disk also alter spinal biomechanics. Chemonucleolysis of the canine disk induces temporary decreased stiffness in lateral bending and permanent reduction of stiffness in flexion and torsion.[77]

The contribution of the various ligaments of the vertebral motion unit to spinal stability in man have been determined by biomechanical testing and comparison of collagen organization and structure.[79] Other studies have used polarized light microscopy, scanning electron microscopy, and x-ray diffraction. It has been demonstrated that stiffness of a ligament is dependent upon the directional orientation of collagen fibers and direction of load.[79] From this data some researchers have concluded that the supraspinous and interspinous ligaments do not function to resist flexion, but rather to cushion the posterior spine. This contradicts theories regarding the posterior elements' contributions to spinal stability,[79] and have been confirmed by VMU range of motion studies that showed no increase in flexion subsequent to severance of the supraspinous and interspinous ligaments.[80] Removal of the supraspinous and interspinous ligaments alone causes a significant decrease of stiffness in flexion.[2] In man the posterior and anterior longitudinal ligaments have been shown to be loaded primarily in flexion and extension, respectively; however, they carry little to no load during lateral bending and torsion. [81]

Surgical alteration of the posterior elements of the lumbar spine may be indicated in human medicine for the relief of lower back pain. Removal of the articular facets may be necessary for exposure of affected nerve roots.[82] The articular facets of the human VMU have been shown to contribute to load transfer in axial, torsional, and flexion-extension loading.[54] Removal of the posterior elements causes an increase in the range of motion in extension.[68] Abumi has reported that the range of motion and neutral zone is significantly altered in axial rotation by lumbar unilateral facetectomy alone and suggested that this is an indication of subsequent instability; however, lateral bending and extension were not significantly affected.[80] The neutral zone is defined as "the region of high flexibility or laxity around the neutral position" of the spine.[83] It differs from the range of motion in that it does not incorporate the elastic zone of bending. In the same study it was demonstrated that a medial facetectomy may be performed without any significant effect on stability. The effect of removal of posterior elements of the human thoracic and lumbar spine has been shown to cause significant decreases in stiffness in shear, bending, and rotation in *in vitro* studies.[4, 57] A two level facet-sparing dorsal laminectomy of the thoracic spine caused a significant decrease in strength in compression.[84] Posterior element removal in the lumbar spine leads to changes of load bearing characteristics and increases in interdiskal pressure although significant increases in mechanical instability seem to require posterior element defects in addition to compromise to the posterior stabilizing elements.[85-87] This evidence suggests that the posterior elements of the spine play a significant role in load bearing particularly in the axial plane and that forces are transferred eccentrically.

Unilateral facetectomy induces a decreased load on the contralateral facet while both unilateral and bilateral facetectomies cause increases in the range of motion in flexion;[88, 89] however, the addition of disruption of the posterior ligaments, including the yellow, interspinous, and supraspinous ligaments, was necessary to cause a significant difference in deflection during flexion-compression loading.[82] The addition of a partial diskectomy to disruption of the posterior ligaments caused a decrease in stability. Results of these studies have suggested that unilateral or bilateral facetectomy does not induce a significant change in the force displacement curve for the described mode of loading; however, alterations in centers of rotation and biomechanical function of the remaining structural elements may lead to degenerative changes including bone remodeling of the remaining joints. In the human spine, the facet joints are considered a major determinant of segmental motion and their alteration may cause pathologic changes that include segmental degeneration, disk herniation, and back pain.[82] They carry significant loads during all directions of loading with the exception of flexion. [81]

The articular facets contribute to load bearing and stability in the canine spine in a similar fashion to facets of the human spine. The load-contact points and relative motion of lumbar facets have been determined in *in vitro* and *in vivo* canine spines.[90-92] Removal of the facet joints, dorsal longitudinal ligament, ligamentum flavum and dorsal and interspinous ligaments significantly alters spinal stability in axial loading and rotation.[63] However, bilateral facetectomy alone has not been shown to cause a significant decrease in stiffness.[1] Hemilaminectomy does not significantly decrease the rigidity of the lumbar spine although it does induce a significant increase in the range of motion in flexion and extension. Bilateral facetectomy results in a significant increase in range of motion and decrease of stiffness in extension but a nonsignificant decrease of stiffness in flexion.[2] Dorsal laminectomy causes a significant decrease of stiffness in flexion and extension and a significant increase in range of motion.

Alterations of the anterior or posterior elements of the human spine cause movement of the instantaneous axis of rotation of the VMU. The migration of the axis distributes loads to the remaining structures which may preserve stability and alters motion coupling in response to loads.[93] Knowledge of the contributions of the anterior and posterior elements of the VMU to spinal stability led to development of the two compartment and three compartment models for spinal stability which have been described and adapted for canine spinal biomechanics. Evaluation of the individual columns is then used to determine the stability of an injured spine. Unfortunately these theories often fail to accurately depict clinical biomechanical status.[68]

The contribution of the paraspinal musculature to human spinal stability has been investigated by determination of muscle composition, strength, and myoelectric activity. Paraspinal muscles contribute to physiologic movement of the spine and may contribute to spinal stability especially in the cervical and lower lumbar regions. The role of paraspinal muscles in spinal trauma is debated. Contraction may be protective by limiting hyperextension and hyper flexion; however some mechanical modeling has demonstrated that muscle contraction may increase injury in cases of axial loading.[68] In three point bending, it is thought that the paraspinal musculature allows the spine to withstand greater loads by altering stress distributions. In lateral bending, multisegmental muscles, particularly those inserting on the pelvis, provide greater stability than intersegmental muscles.[3]

Assimilation of the vast amount of information regarding human spinal biomechanics and application of this information to *in vivo* injury is difficult. One theory that aids in the application is the three subsystem concept presented by Panjabi. The spinal unit is divided into the passive system of the bones and ligaments, the active subsystem of the surrounding muscles and tendons, and the neural subsystem which monitors and directs the other two systems to provide necessary stability.[94] Compromise of one of the subsystems may result in immediate compensation by the other two subsystems, long term adaptation by the remaining subsystems, or significant failure and injury. This model has been proposed as an aid in the diagnosis of spinal instability, particularly in cases of lower back pain in humans. [94] The three subsystem model was proposed during investigations into the neutral zones of the spine in various modes of bending. The neutral zone has been shown to increase with spinal or muscular injury and has been suggested as an index for clinical instability. [83] The neutral zone appears to be a more sensitive indicator of spinal instability in examinations of disk degeneration, spinal injury, and muscular compromise. Insight into the significant value of this new parameter in biomechanics led Panjabi to report the following definition of spinal instability: "Clinical instability is defined as a significant decrease in the capacity of the stabilizing system of the spine to maintain the intervertebral neutral zones within the physiologic limits so that there is no neurological dysfunction, no major deformity, and no incapacitating pain." [83]

II. INTRODUCTION

The functions of the canine vertebral column include protection of the spinal cord, support of the head, support of the thorax and abdomen, and transmission of forces generated by the limbs. This requires flexibility within a functional range of motion with simultaneous maintenance of stability of the spine. Pathologic processes and surgical intervention may result in alterations of the biomechanical properties of the spine through increases or decreases in the range of motion or stability of the spine. A decrease in range of motion between two adjacent vertebrae caused by arthrodesis or ankylosis appears to be well tolerated clinically without alterations to the functions of the spine; however, a decrease of intervertebral stability resulting from pathologic changes or surgical alterations can result in catastrophic spinal cord injury. [4]

Numerous surgical procedures have been described in the dog to expose the spinal canal for the relief of compressive lesions and for diagnosis and therapy of neoplasia. [6, 9, 14, 20] Disruption of the vertebrae and surrounding soft tissue structures by surgical procedures alters the biomechanical integrity of the spine;[2] however, the clinical relevance of these alterations is uncertain. Similarly, spinal column trauma or pathology may result in alterations of spinal biomechanics, but it remains difficult to determine whether specific injuries indicate a need for surgical intervention. To help determine the effect of various surgical procedures and trauma on the spinal column, *in vitro* biomechanical studies can be employed using a servohydraulic testing apparatus and cadaver vertebral motion units.

The vertebral motion unit (VMU) is defined as the smallest functional unit of the spine. It includes two adjacent vertebrae with their interconnecting ligaments, joint capsules, and intervertebral disk. The normal or altered VMU may be subjected to bending, torsional, compressive, or distractive forces by the servohydraulic tester which simultaneously alters the VMU and records the resistant forces. Comparisons may then be made between the range of motion, load to failure, and slope of the load vs. displacement curve of various VMUs.

The effects of various surgical procedures on the flexion, extension, and torsional stiffness of the thoracolumbar spine have been previously investigated.[1,2] This study investigates the effect of surgical alteration of anatomic components of the VMU on lateral bending in the normal canine thoracolumbar spine.

III. MATERIALS AND METHODS

Fresh frozen thoracolumbar spines of 48 clinically mature mix breed dogs weighing between 10 and 25 kilograms were obtained and stored at -20 C. All dogs had been euthanized for reasons unrelated to this research with intravenous pentobarbital overdose. None had a clinical history of spinal disease or trauma. The T_{13}-L_1 vertebral motion units were dissected free of surrounding musculature while preserving the interconnecting ligaments, joint capsules, and intervertebral disk. The cranial and caudal ends of the vertebral bodies of the VMUs were cross-pinned using 0.065 orthopedic pins and mounted in 1 cm thick blocks of polymethymethacrylate molded to fit the testing jig (fig 5). The cross pins served to reinforce the bond between the polymethylmethacrylate and vertebral body. The polymethylmethacrylate was shaped to fit snugly into the testing jig by plastic molds and the VMUs were centered in the mold to standardize the center of rotation. The polymethylmethacrylate blocks were secured into the receptacles of the bending jig with 1/4 inch aluminum block plates and 3 machine screws.

The spines were divided into eight groups of 6 dogs each using a modified blocking technique to eliminate the effect of weight on the comparison of means (Table 1). Group N specimens were not anatomically altered and served as unaltered controls. Group D specimens underwent diskectomy by sharp, full thickness, circumferential incision of the annulus fibrosus with a scalpel blade. Specimens in group F1 were modified by unilateral facetectomies on the right side and specimens in Group F2 were modified by bilateral facetectomies. Facetectomies included complete removal of the cranial and caudal facets, joint capsule, and surrounding bone using rongeurs. Specimens in group FL and FV were modified by right lateral and ventral disk fenestrations respectively. Disk fenestrations were performed by creation of a window in the lateral or ventral annulus with a scalpel blade followed by removal of the nucleus pulposus with a tartar scraper.

Group DF1 and DF2 specimens underwent diskectomies and right unilateral and bilateral facetectomies respectively. Specimens in groups F1FL and F1FV were altered by unilateral right facetectomies and a right lateral (F1FL) or a ventral annulus fenestration (F1FV). Specimens were sealed in plastic and stored at -20 C in between manipulations. The specimens were removed from freezing 5 times for dissection, pinning, potting, surgical manipulating and testing. Exposure to room temperature and humidity was limited to 15 minutes. Prior to testing specimens were thawed to room temperature.

Specimens were tested using an Instron mechanical tester and a swing arm bending jig constructed of 2023-T6 aluminum (fig 6). The jig was designed to produce four point bending (fig 11). Based on the dimensions of the jig, angular displacement (Θ) of the specimen was calculated as Θ = inverse tangent [displacement(inches)/2"]. Distraction of the jig by the servohydraulic materials tester caused vertical bending of the subject material with equal force distribution across the subject. To test lateral bending, the specimens were positioned to produce left sided compression and right sided distraction. A similar bending jig has been previously validated to simulate four point bending within a bending range of 18 degrees.[2] The specimens were positioned within the jig for left sided lateral bending and loaded at a rate of 2.5 cm per minute to failure. The total time for testing each specimen was approximately 5 minutes.

A bending force vs. displacement curve was recorded by a deflection graphic plotter (fig 7) and the mode and site of failure was visually determined and described. Slopes of the initial force vs. displacement curve were calculated after allowing for initial loading of the apparatus. The slope was taken from the linear portion of the graph after the slippage translation was over, and before the plastic deformation began. These results were converted to ratios of moment to angular displacement in newtonmeters (Nm) per degree (Deg) for each specimen. Load and angle at ultimate failure were also recorded. Ultimate failure was determined as the first major negative deflection in the force vs. displacement curve. These values were not recorded if load was not on the scale of the recorder or if a definite point of failure could not be identified.

Differences among mean values of the slopes calculated for the 8 types of surgical alterations were determined by the method of least squares ($p < 0.05$). Analysis of covariance determined that spine weight significantly affected spine stiffness; the least square means adjusted for this effect.

IV. RESULTS

Observation of the bending apparatus during initial loading of each specimen demonstrated minimal shifting of the specimen within the bending jig as the polymethymethacrylate mounting blocks shifted in the connecting blocks of the jig. This shifting was reflected by an initial bending moment vs.

angular deformation curve showing a shallow slope. This initial portion of the curve was disregarded since it did not reflect the actual properties of the vertebral motion unit.

Results of weight, slope, load at failure and angle of deformation at failure are reported by group in tables 2a-2h. Specimens in the control group (N) demonstrated a mean slope of 4.45 Nm/deg (2.9 - 6.4). Specimens undergoing unilateral facetectomies (F1) had a mean slope of 3.50 Nm/deg (2.7 - 4.0) and bilateral facetectomy (F2) decreased the mean slope to 3.28 Nm/deg (2.2 - 4.9). Specimens in the F1FL and F1FV produced mean slopes of 4.26 Nm/deg (3.6 - 5.2) and 3.98 Nm/deg (2.0 - 5.0) respectively. Diskectomy alone (D) caused a decrease in mean slope to 1.70 Nm/deg (0.8 - 2.7) while the addition of unilateral and bilateral fenestration decreased the mean slope to 1.55 Nm/deg (0.4 - 2.3) and 0.86 Nm/deg (0.3 - 1.6) respectively.(Table 1)

Catastrophic failure of all specimens in groups N, F1, F2, F1FL, and F1FV occurred primarily at one of the vertebral body end plates with simultaneous disruption of the T13-L1 intertransverse ligament. Failure of specimens in groups undergoing diskectomies alone occurred at the contralateral facet joint capsule. Specimens modified by a combination of diskectomy and facetectomy failed by disruption of the indistinct ligamentous tissue between the thirteenth rib and the transverse process.

A statistical difference was found between the stiffness of all diskectomy groups when compared to any other group. The addition of unilateral facetectomy or bilateral facetectomy to the diskectomy model did not significantly decrease the stiffness of the model. The decrease in stiffness caused by unilateral or bilateral facetectomy alone was also not significant when compared to the control. Blocking by weight eliminated the effect of weight on treatment; however, there was a significant correlation between body weight and stiffness (fig 9). Vertebral motion units harvested from larger dogs demonstrated a significant increase in stiffness over those of smaller dogs.

V. DISCUSSION

In vitro servohydraulic testing has been used extensively in investigations of human and canine spinal biomechanics in spite of its limitations. These limitations include eliminating the effects of the paraspinal musculature on spinal stiffness, and the properties of the testing design which limits evaluation of biomechanical characteristics of the normal spine. For example, the geometry of the vertebrae and surrounding soft tissues cause motion coupling in nearly all movements of the normal spine. Coupling is the simultaneous movement in one degree of freedom that is dictated by movement of the spine within another degree of freedom. Biomechanical testing may eliminate coupling, thereby providing an inaccurate model for overall spinal motion studies. Similarly, this and other studies use single episodes of force applied to failure that ignores viscoelastic properties of the spine.

Freezing at -20 C has been shown to have limited effects on the biomechanics of the bone, ligaments and disk of the vertebral motion unit.[95-97] Although freezing and thawing of VMUs for future testing is routinely employed, the specific effects of multiple environmental changes on the spinal model are unknown. The ideal condition for manipulations at room temperature is 100% humidity although no significant alterations in biomechanical properties have been recognized after 10 minutes in normal room conditions.

The greatest limitation to application of *in vitro* spinal biomechanics to clinical situations may be the difficulty in defining spinal instability. Clinically, instability may be defined as the "the loss of the ability of the spine under physiologic loads to maintain its pattern of displacement so that there is no initial or additional neurologic deficit, no major deformity, and no incapacitating pain."[5] Spinal biomechanics, however, represents a continuum and, therefore, identification of a specific point of instability becomes difficult. Recently, Panjabi has proposed the neutral zone as a sensitive indicator of spinal instability in *in vitro* and *in vivo* spinal models. The neutral zone is "the region of high flexibility or laxity around the neutral position." It differs from the range of motion in that it does not incorporate the elastic zone of bending.[83]

This study demonstrates that lateral or ventral disk fenestration in addition to unilateral or bilateral facetectomies do not significantly alter the stiffness of the thoracolumbar VMU in lateral bending as reflected by the load vs. deformation curve. This anatomic region was selected because of the high incidence of decompressive procedures and traumatic injuries reported in these locations. The cause of higher slopes among specimens undergoing fenestration and unilateral facetectomy relative to those altered by unilateral or bilateral facetectomy is unknown. Because these differences were not statistically significant it is likely that they represent normal variation secondary to weight, age, breed or sex.

Although the significant effect of weight on spinal stiffness was eliminated by blocking, the effects of age, sex and breed were not explored. Age has been shown to have a significant effect on spinal stiffness over time [78] and may have altered the results of this investigation; however, the effects of breed and sex on spinal stability are uncertain. The results may also have been confounded by previous spinal disease that effected the spinal biomechanics of the specimens. Radiographic examinations of the spines may have demonstrated pathologic abnormalities that were not suspected based on clinical history.

The increased stiffness associated with increased weight is most likely due to the larger size of the vertebral motion units of heavier animals. This increase in size allows a greater mechanical advantage by the lengthening moment arms of the vertebral body, subsequently providing greater resistance to bending. Although the material properties of the vertebral motion units of all normal dogs are probably similar, the increased volume of stabilizing materials such as ligaments and disks should also increase the stiffness of the spines of larger dogs.

Clinical evaluation of the anatomy and posture of the dog would suggest that there is a more limited range of motion and greater stiffness of the spine in lateral bending than in dorsal and ventral flexion; however, the mean slope of moment to angular deformation in lateral bending was approximately 30% of that determined in flexion studies.[2] This may be due to differences in experimental techniques, particularly in the methods of determining deformation.

Deformation in this study was recorded from the servohydraulic output. This reflects the change in distance of the Instron clamps which is separated from the VMU by the testing jig and mounting apparatus. Slippage associated with these parts of the testing apparatus may be inappropriately interpreted as deformation of the VMU, falsely lowering the calculated stiffness. Determination of deformation in Smith's study was performed by means of a rotatory variable differential transformer attached directly to the VMU. This technique eliminates the effects of the mounting apparatus on the measurement of deflection and the addition of this apparatus probably adds little resistance to the VMU. The mounting technique of vertebral body pinning used in Smith's study may also have been more stable than the polymethylmethacrylate potting used in this study, but the length of the spinal segment used in that study made the fixation technique possible.

Standardization of mounting and testing techniques used in spinal biomechanical testing would make the comparison of data between studies more accurate. Similarly, agreement on which biomechanical parameters are most clinically significant and most valuable for reporting would also aid in comparisons between studies. Possible measurements include load to deformation ratios, load and deformation at failure and range of motion. Range of motion was not determined in this study because of the difficulty of accurate determination of changes in the relatively limited range of motion in lateral bending of the thoracolumbar spine.

Results of testing in lateral bending suggest that the articular facets of the canine thoracolumbar spine are more important to stability in axial rotation and dorso-flexion than in lateral bending and ventro-flexion. This is supported by strain gauge testing of canine lumbar facet joints which demonstrated larger loads in torsion and extension than in flexion or lateral bending.[91] Bilateral facetectomy did not cause a significant decrease of stiffness in rotational studies of thoracolumbar vertebral motion units.[1] Adding ventral or lateral fenestration to a unilateral facetectomy did lead to significant decreases in stiffness. Unilateral facetectomy in combination with diskectomy significantly reduces torsional stiffness but diskectomy alone does not.[1] Hemilaminectomy alone did not alter the stiffness of canine L_3-L_4 VMU in flexion and extension bending studies.[2] Bilateral facetectomy resulted in a significant increase in range of motion and a decrease of stiffness in extension.[2]

These findings suggest that unilateral facetectomy may be performed safely without causing instability of the canine spine. While a combination of unilateral facetectomy and fenestration decreases the stability of lumbar vertebral motion units in axial rotation, this may not be of clinical significance when the paraspinal musculature is neurologically and functionally intact and can provide additional stability.

Failure of the test specimens in the non-diskectomy groups at the cartilaginous end plates is consistent with findings regarding the relative strength of the intervertebral disk, vertebral body, and cartilaginous endplate. Axial tension applied to human lumbar spines most frequently resulted in failure at the cartilaginous endplate versus the annulus fibrosus or vertebral body.[66, 71] Diskectomy has been previously used as a model of vertebral body fractures in biomechanical testing based on the relative strengths of the annulus and vertebral body.[1, 98, 99] The lumbar spine does not demonstrate a significant decrease in torsional stiffness after diskectomy;[1] however, the significant decrease in lateral bending stiffness induced by diskectomy in our study suggests that dogs with vertebral body fractures may benefit from surgical stabilization.

Clinical determination of spinal stability subsequent to trauma, disease, or surgical intervention should ultimately be based on evaluation of the entire spinal unit. The spinal unit has been divided into the passive system of the bones and ligaments, the active subsystem of the surrounding musculature and tendons, and the neural subsystem which monitors and directs the other two systems to provide necessary stability.[94] Knowledge of biomechanical properties of the appropriate vertebral motion unit and the functional integrity of the surrounding musculature and its nervous innervation will aid surgeons in determining when additional spinal stabilization is indicated.

The results of this study are difficult to weight because of the variability imposed by the model, the testing method and the specimen variations. Despite these scientific drawbacks the information gleaned was consistent enough to be able to draw some limited conclusions regarding the biomechanical behavior of a T13-L1 VMU in lateral bending. The effects of routine surgeries are limited enough to support the continued use of such procedures on a routine basis. The most significant effects of the spinal fracture model support a concerted effort to further study fixation methods with renewed vigor.

TABLE 1.
MEAN SLOPE, LOAD AT FAILURE AND ANGLE AT FAILURE BY TREATMENT GROUP

Group	Alteration	Mean Slope	Mean Load at Failure	Mean Angle at Failure
N	Normal control	4.45 Nm/Deg	74 Nm	14 Deg
F1	Unilateral facetectomy	3.50 Nm/Deg	55 Nm	14 Deg
F2	Bilateral facetectomy	3.28 Nm/Deg	52 Nm	15 Deg
F1FL	Facetectomy/lateral fenestration	4.26 Nm/Deg	46 Nm	11 Deg
F1FV	Facetectomy/ventral fenestration	3.98 Nm/Deg	58 Nm	13 Deg
D	Diskectomy	1.70 Nm/Deg	19 Nm	14 Deg
DF1	Diskectomy/unilateral facetectomy	1.55 Nm/Deg	15 Nm	15 Deg
DF2	Diskectomy/bilateral facetectomy	0.86 Nm/Deg	7.2 Nm	13 Deg

TABLE 2a.
RESULTS OF TESTING IN LATERAL BENDING
NORMAL CONTROL SPECIMENS (N)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
1	10.4	4.4	102	20.0
2	11.4	3.0	60	11.3
3	11.4	4.7	--	----
4	12.7	2.9	45	11.3
5	17.3	6.4	60	8.5
6	22.7	5.6	102	20

TABLE 2b.
RESULTS OF TESTING IN LATERAL BENDING
UNILATERAL FACETECTOMY (F1)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
7	10.9	3.6	36	9.9
8	13.6	3.4	25	7.1
9	15	3.6	36	14
10	15.9	3.7	83	18
11	19.1	2.7	36	14
12	20.9	4.0	111	20

TABLE 2c.
RESULTS OF TESTING IN LATERAL BENDING BILATERAL FACETECTOMY (F2)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
13	10.0	2.2	30	14
14	13.2	2.6	39	11.3
15	13.4	3.2	55	16.7
16	15.9	2.8	75	19.3
17	16.8	4.5	60	14
18	18.2	4.9	--	--

TABLE 2d.
RESULTS OF TESTING IN LATERAL BENDING
UNILATERAL FACETECTOMY/LATERAL FENESTRATION (F1FL)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
19	10.0	3.6	33	8.5
20	13.6	3.9	60	14
21	15.4	2.9	24	8.5
22	18.2	3.9	33	9.9
23	19.5	4.4	72	16.7
24	22.7	5.2	53	9.9

TABLE 2e.
RESULTS OF TESTING IN LATERAL BENDING
UNILATERAL FACETECTOMY/VENTRAL FENESTRATION (F1FV)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
25	10.0	2.0	14	7.1
26	10.0	5.0	17	7.1
27	16.4	4.4	55	14
28	17.7	4.9	75	16.7
29	18.2	4.9	78	15.3
30	24.1	4.4	111	18

TABLE 2f.
RESULTS OF TESTING IN LATERAL BENDING
DISKECTOMY (D)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
31	9.1	0.8	9.0	16.7
32	12.7	1.4	13	7.1
33	15.9	1.6	16	5.7
34	18.2	2.2	27	22
35	18.2	2.7	30	14
36	20.9	1.5	16	18

TABLE 2g.
RESULTS OF TESTING IN LATERAL BENDING DISKECTOMY/UNILATERAL FENESTRATION (DF1)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
37	11.4	0.4	5.5	9.9
38	11.4	0.7	9	22
39	18.2	1.9	15	16.7
40	18.2	2.3	30	22
41	22.7	2.0	20	9.9
42	27.3	2.0	10	9.9

TABLE 2h.
RESULTS OF TESTING IN LATERAL BENDING DISKECTOMY/BILATERAL FACETECTOMY (DF2)

DOG #	WEIGHT KG	SLOPE NM/DEG	LOAD AT FAILURE NM	ANGLE AT FAILURE DEGREES
43	9.5	0.3	--	--
44	13.6	1.0	3.3	9.9
45	13.6	1.1	6.7	15.6
46	13.6	1.6	15	18
47	16.4	0.6	4.5	7.1
48	19.5	0.6	6.7	16.7

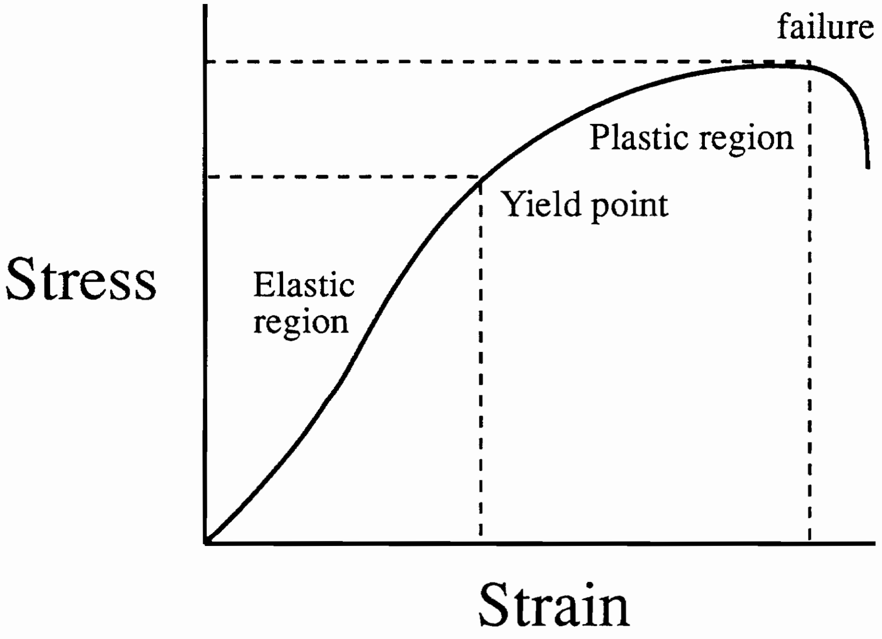


Figure 1. Typical Stress Strain Curve.
Response of an elastic material to load.
Stress = applied load
Strain = resultant deformation

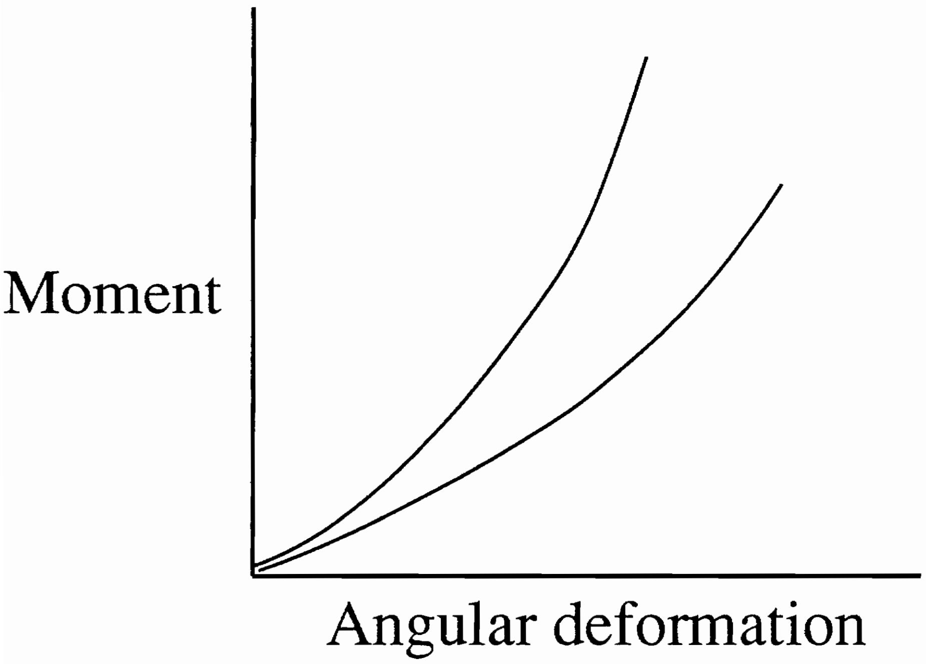
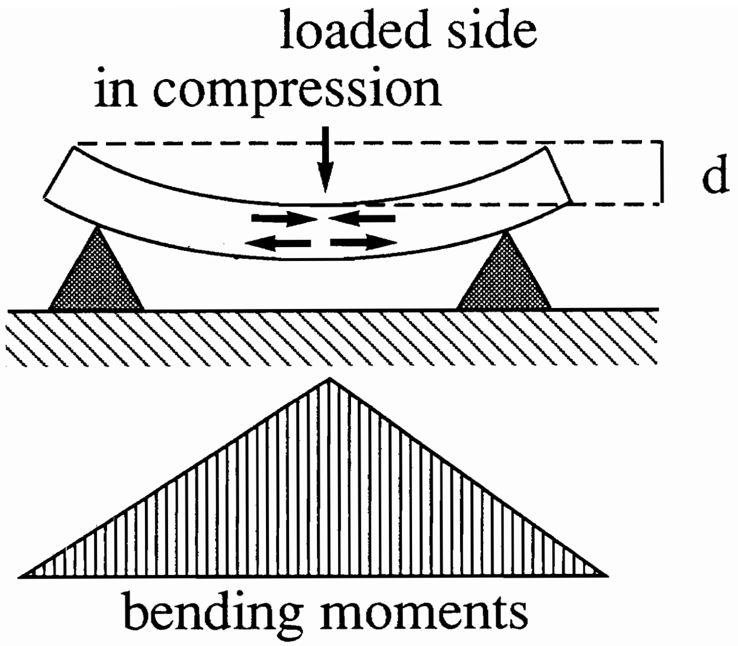


Figure 2. Load deformation curves



Three Point Bending

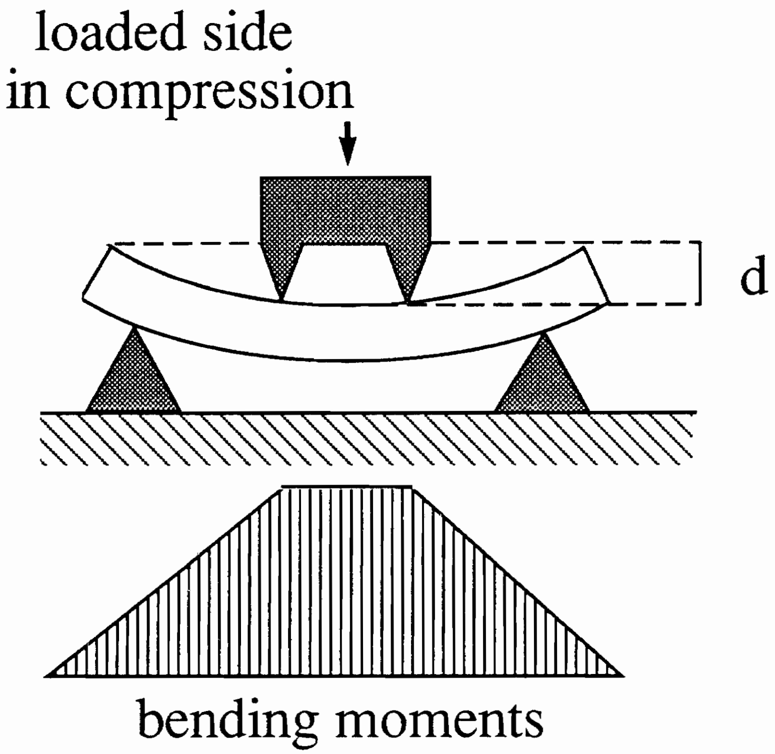
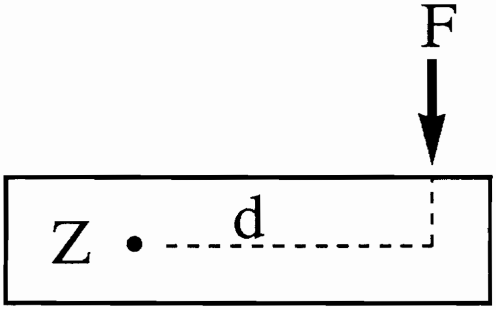


Figure 3. Four Point Bending



$$M_Z = d \times F$$

Figure 4. Moment arm. F = force, d = distance, Z = center of rotation, M_z = moment at Z

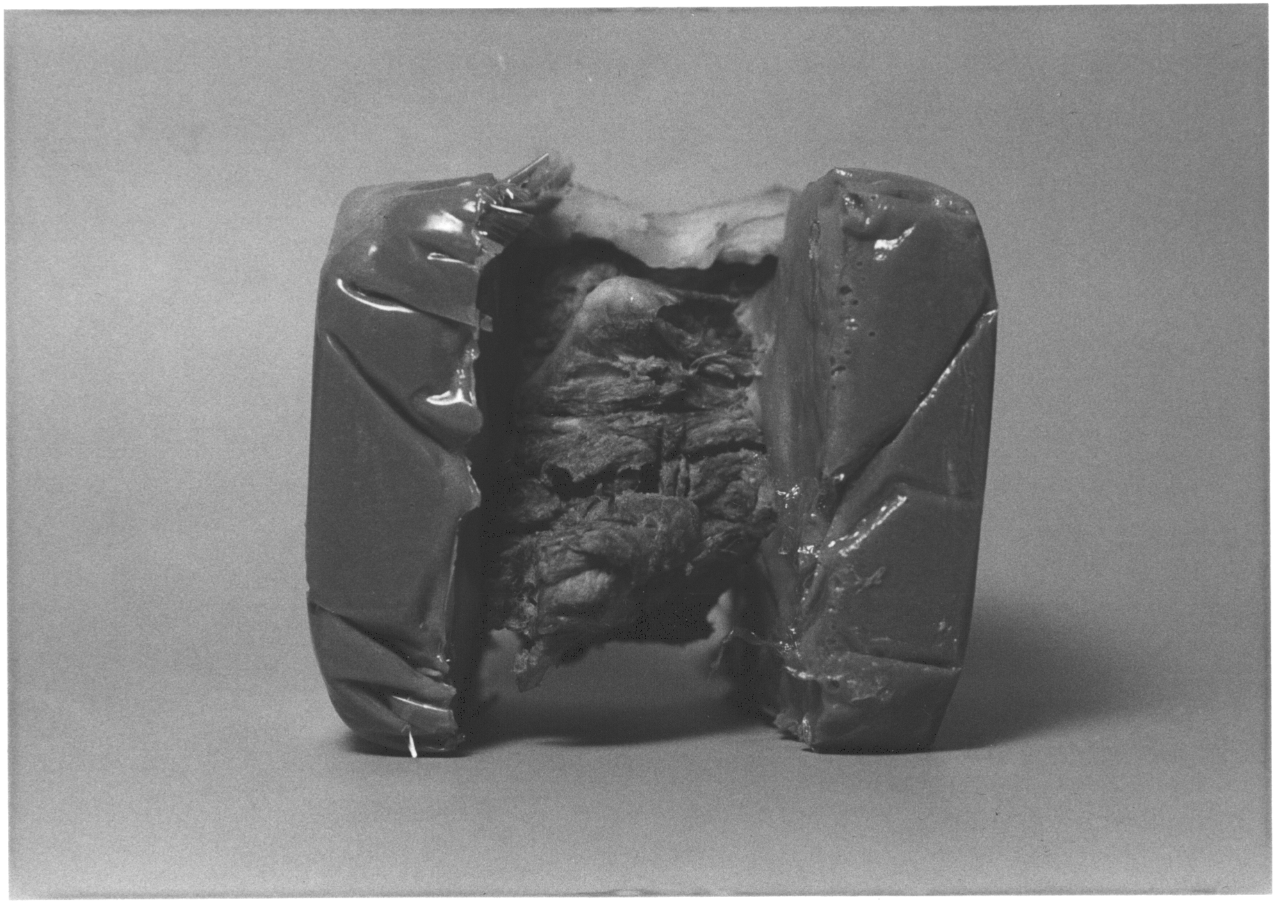


Figure 5. Thoracolumbar vertebral motion unit prepared for testing.

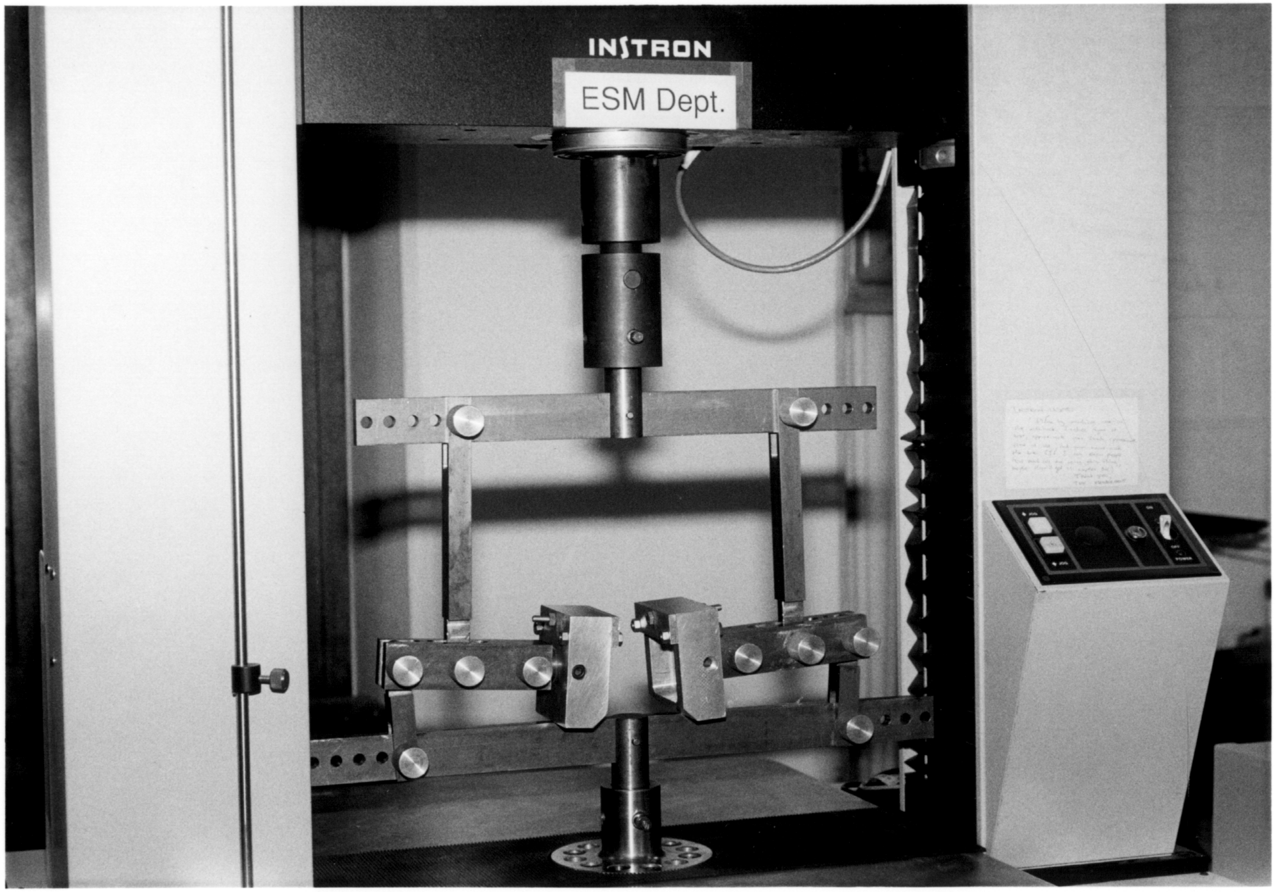


Figure 6. Instron servohydraulic materials tester with four point simulation jig.

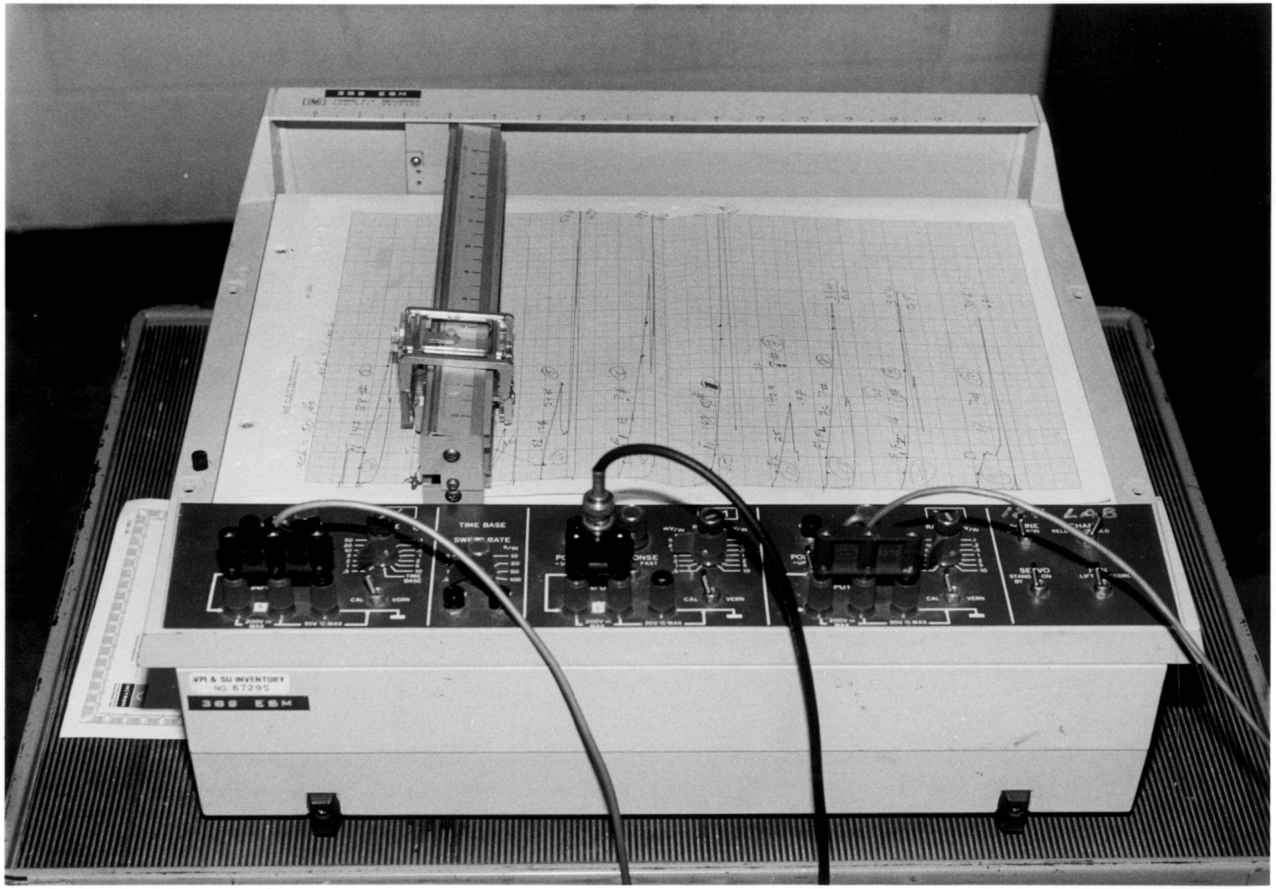


Figure 7. Deflection graphic plotter and results graph

Figure 8. Thoracolumbar vertebral motion unit after testing to failure

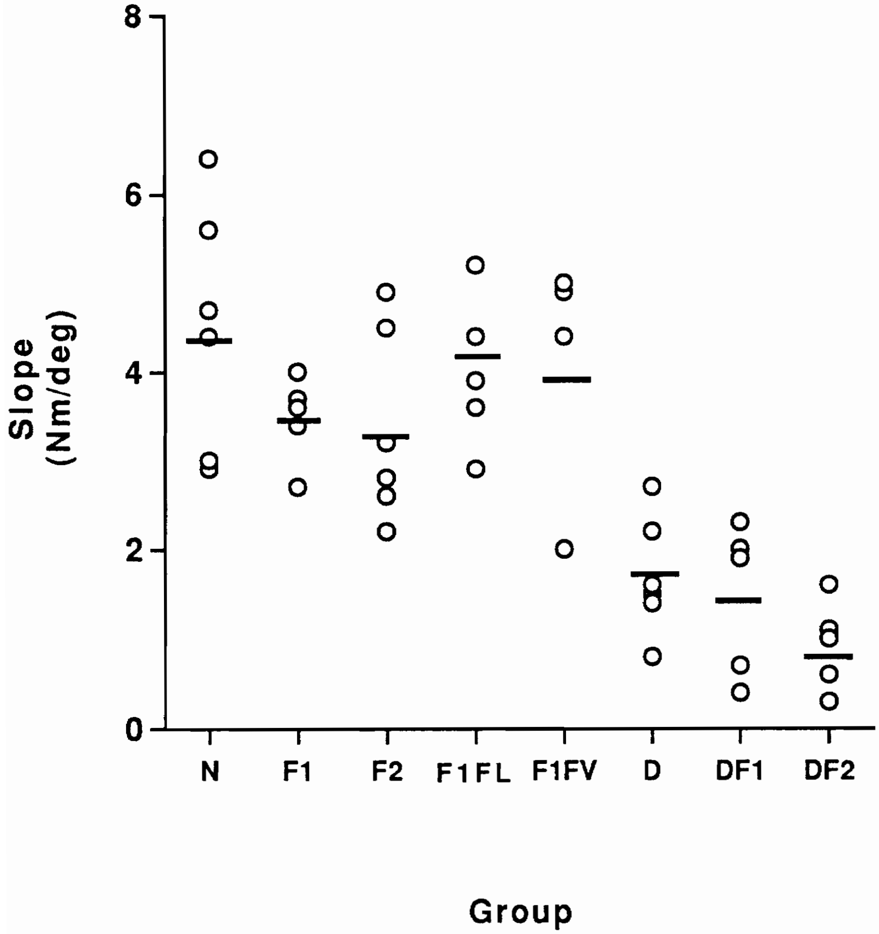


Figure 9. Test Group versus Slope

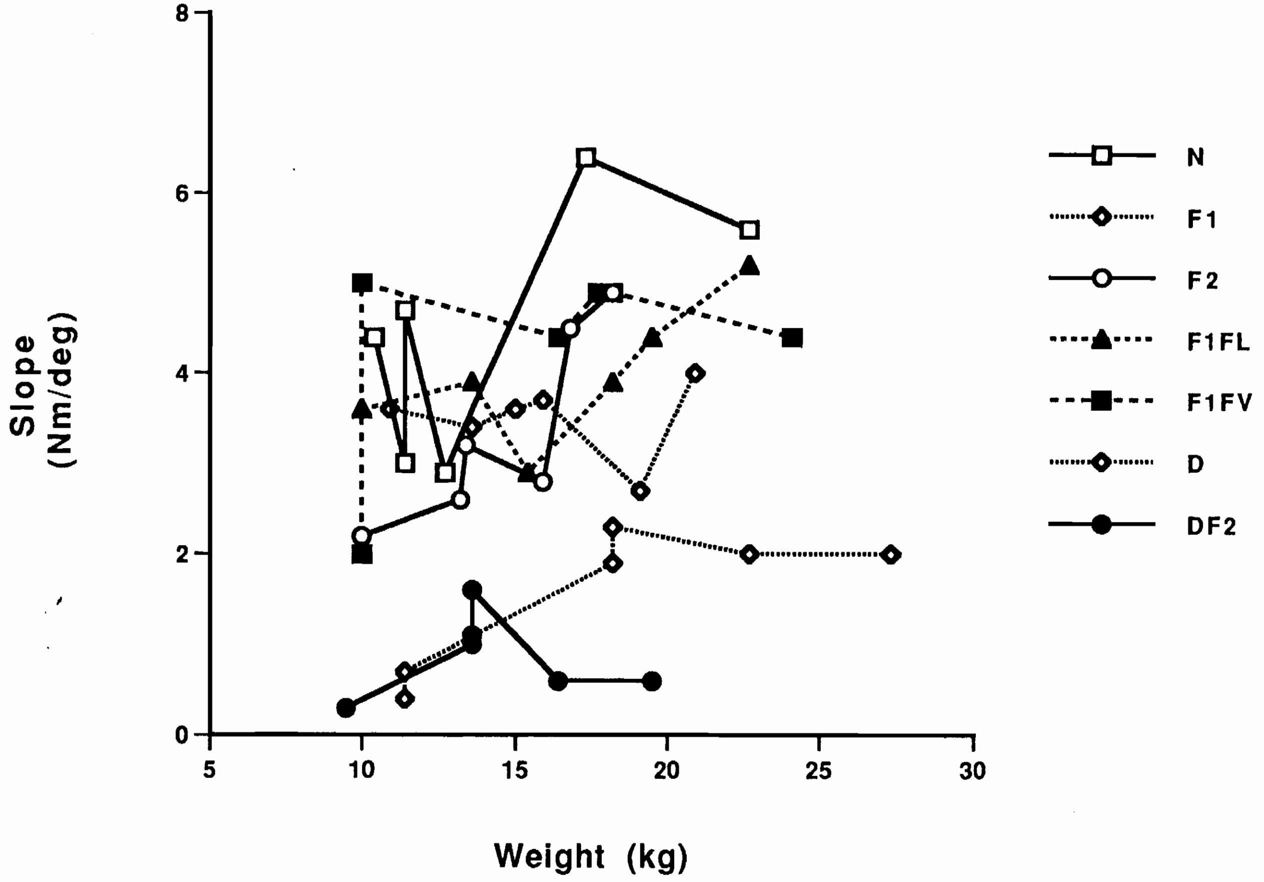


Figure 10. Body Weight versus Stiffness

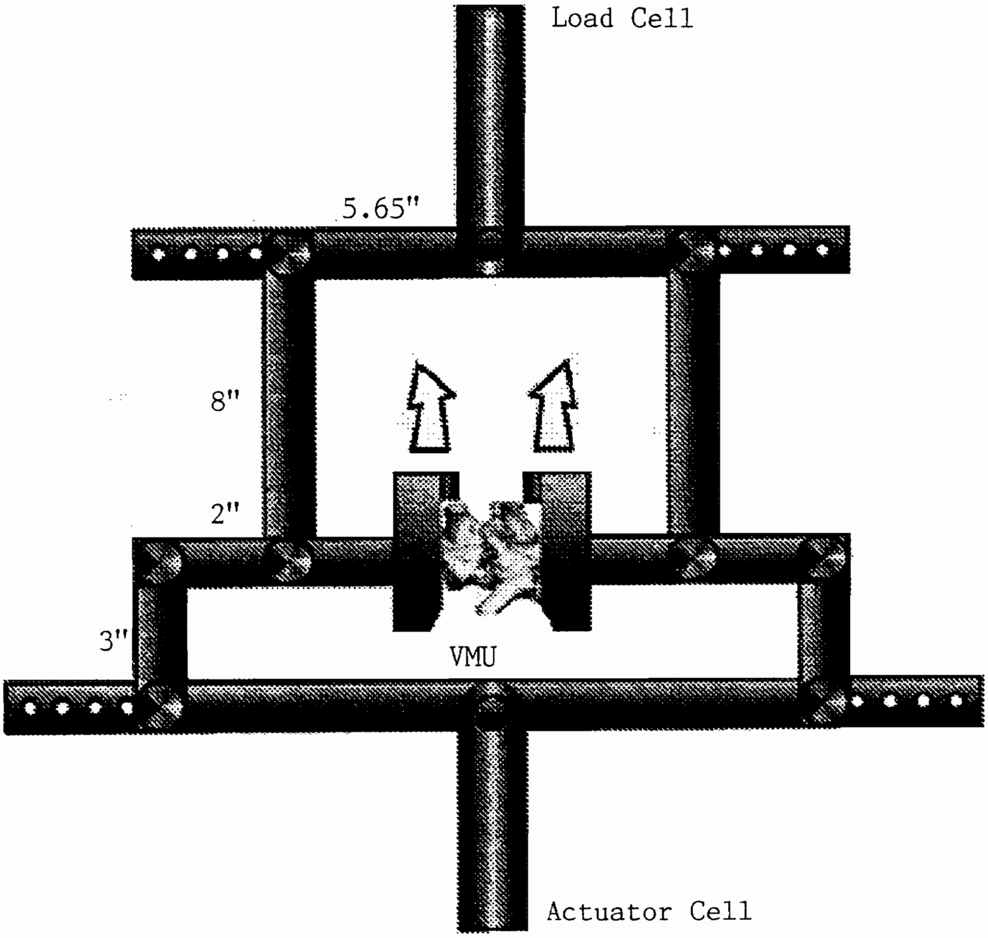


Figure 11. Diagram of Testing Jig with VMU

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VII. VITA

Kurt Sanderson Schulz was born to Joan and Allan Schulz on December 3, 1962. Kurt completed high school with tremendous assistance from his grandmother, Eleanor Russell, and then spent a leisurely year in Rauma, Finland as a Rotary Club exchange student. He returned, relaxed and ready to attend Colgate University in Hamilton, New York. He was appointed to the athletic affairs committee after serving as captain of the Colgate rowing team for two years. Kurt graduated from Colgate, Magna Cum Laude, Phi Beta Kappa, with a Bachelor of Arts in biochemistry in 1985 and moved fifty miles west to Ithaca, New York to pursue a D.V.M. at Cornell University. While attending Cornell Kurt served as treasurer of the open house committee, large animal crew member, director of the junior class skit, published his first case report on horses and, most importantly, served as social chairman of Omega Tau Sigma. Kurt's summers were spent T.B. testing cows in the back woods of New York state and driving busses in the Rocky Mountains of Glacier Park, Montana. Amidst these other activities, he completed his Doctor of Veterinary Medicine degree in 1989. With prodding from Drs Sharon Center DACVIM and Susan Daugherty DACVIM Kurt accepted an internship in small animal medicine and surgery at the University of Missouri, Columbia. After completion of a wonderful and educational internship in mid-Missouri, he reverted to his large animal ways by joining the Peace Corps' veterinary division in Boulmalne Dades, Morocco. While practicing primitive herd health in the Atlas Mountains, Kurt applied for small animal surgery residencies in the United States for the following year. Ranking was performed in-absentia by his mother based purely on proximity to her location. His term of duty was cut short by the Gulf War and he and his partner Dr. Jon Bramson VMD returned home to small animal practice. Kurt began a residency in small animal surgery at the Virginia/Maryland Regional College of Veterinary Medicine in July of 1991. His primary research interests have included neurosurgery and orthopedics.

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