AN IN-VITRO KINEMATIC BIOMECHANICAL EVALUATION OF TWO SURGICAL TECHNIQUES IN ADOLESCENT IDIOPATHIC SCOLIOSIS

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ABSTRACT

The current study was an *in-vitro* kinematic biomechanical evaluation of surgical techniques in adolescent idiopathic scoliosis (AIS). The objective was to compare the three-dimensional range of motion (ROM) of the spinal column between Ponte osteotomy and bilateral medial facetectomy.

AIS is a three-dimensional deformity of the spine that often requires surgical intervention to achieve a balanced, safe, and stable correction. The current destabilization technique involves bilateral medial facetectomy. The difference in the achievable range of motion between facetectomy and Ponte osteotomy has not been quantified previously.

Three-dimensional flexibility tests were conducted on ten thoracolumbar bovine specimens in a repeated measures design with $\alpha = 0.05$ to determine differences in ROM between facetectomy and Ponte osteotomy. Facetectomy entailed bilateral resection of the inferior articular processes, while Ponte osteotomy involved further resection of the supraspinous and interspinous ligaments, spinous process, ligamentum flavum, and bilateral resection of the facet joint and capsule at each surgical level.

Flexibility tests in axial rotation (AR), flexion-extension (FE), flexion-extension with a 660 N compressive follower load (FEFL), and lateral bending (LB) were performed under $\pm 5$ Nm of pure moments for five cycles. Three-dimensional kinematic
data were captured using a high-speed camera, and the overall range of motion (ROM) was determined in each direction.

Compared to the intact state, there was a significantly greater increase in AR ROM after Ponte osteotomy (406% of intact) than after facetectomy (238% of intact). FE trials showed a similar result with a significantly larger increase in ROM after Ponte osteotomy (186% of intact) than after facetectomy (129% of intact). FEFL also yielded a larger increase after Ponte (151% of intact) than after facetectomy (110% of intact). The same was not true of lateral bending, however. In the LB loading condition, the increase in ROM after Ponte osteotomy (117% of intact) was not significantly larger than after facetectomy (113% of intact).

The current study showed that, while bilateral medial facetectomy increased ROM in all three directions, Ponte osteotomy produced a significantly greater increase in ROM in flexion-extension and axial rotation over facetectomy when compared to the intact state.
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CHAPTER I

INTRODUCTION

This project was composed of two distinct segments. The first involved the planning and implementation of an improvement to the existing spine flexibility-testing machine. The second segment involved executing a biomechanical study to quantify and compare three-dimensional range of motion (ROM) achievable after bilateral medial facetectomy (BMF, facetectomy) and after Ponte osteotomy (PO).

1.1 Modification to Spine Flexibility-Testing Machine

The spine flexibility-testing machine (spine machine) (Figure 1.1) in the Biomedical Engineering Department was built by a previous student [1] and has been used to evaluate three-dimensional spinal kinematics by employing a flexibility testing protocol [2]. The torque used to move and bend the spinal specimen was generated by a motor (Model: AKM22E-VBNPC, Danaher Motion Inc., Radford, VA) and actuator arm assembly. Located at the distal end of the actuator arm is a “U-joint” that connects the arm to the specimen. The weight of the actuator arm caused the center of mass to move from the specimen toward the torque arm. In order to balance the weight and return the center of mass to the specimen, a counterbalance was attached to the actuator arm after the connection to the specimen. With the weight balanced on either side of the specimen, the center of mass was returned to the vertical center of the specimen. After restoring the
balance, however, the weights of the actuator arm and counterbalance resulted in a compressive vertical load on the specimen (Figure 1.2).

Figure 1.1 – Spine flexibility-testing machine
The passive XY table, torque (actuator) arm assembly and counterweight are indicated.

Figure 1.2 – Compressive force on specimen due to equipment weight
In order to counteract this load, a counterweight equal to the mass of the compressive load (2.5 kg) was utilized. A thin rope was attached to the proximal end of the specimen. This rope ascended to a horizontal metal bar with a pulley at each end, as shown in Figure 1.3. The rope was brought over one pulley and guided along the metal bar and over the second pulley from which it descended back to the specimen. The rope was then attached to the proximal end of the specimen on the opposite side from the other end of the rope.

A second rope was then attached to the center of the horizontal bar. This rope was then guided over a larger pulley attached to a passive XY slide (Figure 1.4-Figure 1.5). Once over the pulley the free end of the rope was attached to a deadweight. The pulley mechanism and deadweight were thus suspended directly above the test specimen.

During flexibility testing, the counterweight moved passively along the XY table with respect to the test specimen.
In the original design, the low-friction, passive XY slides allowed the counterweight to be aligned vertically above the specimen. The slides were not completely frictionless, however, and the force required to overcome the static friction in the slides resulted in undesirable horizontal loads on the specimen (Figure 1.5) when the ropes became angled. The testing protocol suggested applying a purely rotational moment to the specimen. When a horizontal component was present in the applied load, the moment was no longer pure and was applied unevenly along the specimen. The most distal vertebrae were then subjected to larger loads based on their distance from the loading site.

Figure 1.5 – Depiction of undesired horizontal load applied to specimen
M=applied moment, d=distance (moment arm), $F_x$=horizontal component of force, $F_y$=vertical component of force

In order to reduce the horizontal load component on the specimen, it was necessary to keep the counterweight suspended directly above the specimen thereby maintaining a vertical alignment in the ropes attached to the specimen. In the newly
implemented mechanism, this was accomplished by incorporating an actively controlled counterweight system into the spine machine. In this design, a motorized XY bi-slide followed the movement of the proximal end of the specimen. The lateral forces exerted on the specimen were reduced by keeping the counterweight suspension cables vertical. The bi-slide was controlled manually using a jog command. The bi-slide operator would jog the slide servomotor in the same direction the specimen was moving. When the order reversed, the operator would jog the servomotor in the opposite direction. One slide was used for lateral bending and the other for flexion-extension. The bi-slide did not move for axial rotation since that motion was torsional. The same operator jogged the bi-slide for all testing. Upon completion of the testing machine improvement, the biomechanical study to evaluate the kinematics of the spine comparing bilateral medial facetectomy and Ponte osteotomy commenced.

1.2 Overview of Biomechanical Study

Adolescent idiopathic scoliosis (AIS) is a three-dimensional deformity of the spine. The current destabilization technique involves bilateral medial facetectomy (BMF) and allows for minimal deformity correction.

The Ponte osteotomy (PO) [3] is a posterior-only surgical technique that involves total facetectomy and excision of the ligamentum flavum, interspinous and supraspinous ligaments. The Ponte procedure destabilizes the spine and allows for maximal deformity correction. The difference in the achievable ROM between BMF and PO has not been quantified previously. The surgical efficacy of the Ponte procedure has been alluded to in previous clinical studies [3, 4, 5] but not its biomechanical efficacy.
1.3 Purpose

The purpose of this in-vitro biomechanical investigation was to quantify and compare the achievable three-dimensional range of motion (ROM) of the spinal column between Ponte osteotomy and bilateral medial facetectomy. The flexibility represents the intra-operative amount of deformity correction that is achievable.

1.4 Statement of Clinical Relevance

The Ponte osteotomy, when used to correct AIS, is a relatively novel technique, and the ROM attainable with this procedure has not been quantified or compared previously. The current study determined the amount of scoliosis correction achievable with the Ponte technique, which provides clinicians with critical information in deciding whether or not the additional osteotomy is justified.

1.5 Specific Aims of the Study

The specific aims of the current study are to:

1. Incorporate an actively controlled counterbalance system into an existing spine flexibility-testing machine.

2. Quantify the range of motion achievable between Ponte osteotomy and facetectomy.

1.6 Null Hypotheses

1. There are no differences in the total achievable ROM between facetectomy and Ponte osteotomy for the AR loading condition.
2. There are no differences in the total achievable ROM between facetectomy and Ponte osteotomy for the FE loading condition.

3. There are no differences in the total achievable ROM between facetectomy and Ponte osteotomy for the FEFL loading condition.

4. There are no differences in the total achievable ROM between facetectomy and Ponte osteotomy for the LB loading condition.

1.7 Alternative Hypothesis

1. There are differences in the total achievable ROM between facetectomy and Ponte osteotomy for the AR loading condition.

2. There are differences in the total achievable ROM between facetectomy and Ponte osteotomy for the FE loading condition.

3. There are differences in the total achievable ROM between facetectomy and Ponte osteotomy for the FEFL loading condition.

4. There are differences in the total achievable ROM between facetectomy and Ponte osteotomy for the LB loading condition.
CHAPTER II

LITERATURE REVIEW

2.1 The Human Spine

The human spine is comprised of 33 vertebrae, which are divided into anatomical segments, as shown in Figure 2.1. The uppermost segment consists of seven vertebrae (C1-C7) and is called the cervical spine. Immediately inferior to the cervical spine is the thoracic spine with its 12 vertebrae (T1-T12). Proceeding caudally, the next section

Figure 2.1 – A model of the human spine indicating anatomical regions
contains the five vertebrae of the lumbar spine (L1-L5) which are followed by the sacrum containing five fused vertebrae (S1-S5). The most distal segment is composed of the four fused bodies of the coccyx (Co1-Co4).

Each vertebra is comprised of a vertebral body and its posterior elements (Figure 2.2). These elements include the pedicles (P), transverse processes (TP), superior and inferior articular processes (SAP and IAP, respectively), spinous process (SP), and laminae (L). The inferior articular process of one vertebra fits into the superior articular process of the adjacent vertebra. The diagram illustrates these relationships from various perspectives.

Figure 2.2 – Views of the human vertebral body and posterior elements from various perspectives (Reprinted with permission [40, p. 18])
process of the vertebra immediately above forming the zygapophysial (facet) joint (Figure 2.3). This is the only articulating joint found in the spine.

The spinal vertebrae are stacked vertically and separated by intervertebral discs (IVD) that act as shock absorbers. The biomechanical functions of the spine are to support loads, allow motion, and protect the spinal cord. Similarly, the functions of the IVD and facets are to support loads and to allow or limit motion.

The ligaments of the human spine, shown in Figure 2.4, are the anterior and posterior longitudinal ligaments (ALL and PLL, respectively), ligamentum flavum (LF), interspinous ligament (ISL) and supraspinous ligament (SSL). Their purpose is to assist in providing protection and stability to the spinal column.
2.2 Flexibility Testing

In accordance with the spine flexibility testing protocol [2], a pure bending moment was applied to the proximal end of a multi-segmented specimen while the distal end was rigidly attached to the spine machine platform. A pure bending moment [6] was exerted by the application of a purely rotational load in the absence of simultaneous shear forces. The multi-segmented spine was unconstrained in translation so that the shear forces were zero. Bending moments were used for testing in flexion-extension (FE), lateral bending (LB) and axial rotation (AR) (Figure 2.5). The same spine machine was used to apply pure moments in these directions.
2.3 Coordinate Systems

In order to perform the flexibility testing, both global and local coordinate systems must be established so that the motion has a frame of reference. Wilke, et al. set forth a standardized protocol for determining the coordinate systems [7].

The x-axis is aligned in the anteroposterior direction, and lateral bending is a rotation about the x-axis. The y-axis is aligned mediolaterally, and flexion-extension motion is a rotation about the y-axis. Similarly, the z-axis is aligned in the craniocaudal direction, and axial rotation is a rotation about the z-axis (Figure 2.6).
2.4 Viscoelastic Properties of Spinal Tissues

Among the biomechanical properties of the spine are the viscoelastic properties. These refer to time-dependent deformation characteristics associated with biological tissues. Specimens will attain an equilibrium state over time when a static or dynamic load is applied. The spinal motion segments are known to reach an equilibrium state after three cycles of loading [7].

2.5 Range of Motion

The biomechanical measure most pertinent to this study was range of motion. The spine machine was capable of moving the specimen in six degrees of freedom, including translations along and rotations about three orthogonal axes. The motions of interest in the current study were rotations about the x-axis (lateral bending (LB)), y-axis (flexion-extension (FE and FEFL)), and the z-axis (axial rotation (AR)).

A load-displacement curve with three test cycles is shown in Figure 2.7. Each “S-curve” represents one testing cycle. It can be seen that the first cycle’s ROM is less than

![Figure 2.7 – Typical load-displacement curve (Reprinted with permission [7])](image)
the third, and that with each cycle, the endpoints become increasingly close together until they are nearly the same. This represents the approach of viscoelastic equilibrium as the curve nears a steady state after the third cycle [7].

2.6 Follower Load

A follower load (FL) was used in this experiment during the flexion-extension posture (FEFL). A follower load is a load whose resultant force is directed along the spinal curvature [8]. Studies have shown that the FL stabilizes the spine and allows for increased load-carrying capacity [9, 10]. The drawing in Figure 2.8(A) indicates the direction of a vertical compressive force. Figure 2.8(B) shows the compressive follower load and indicates its direction along the curvature and through the centers of rotation at each vertebral level.

Figure 2.8 – Schematic drawing of follower load configuration
(A) Compressive vertical load, (B) Compressive follower load
2.7 Adolescent Idiopathic Scoliosis

A spinal deformity as described by Harry Shufflebarger, MD, is “any condition which alters the anatomy or functional integrity of the spinal motion segment.” [11] Adolescent idiopathic scoliosis (AIS) is a three-dimensional deformity that involves lateral deviation and axial rotation of the spine [12]. It affects 2% to 2.5% of most populations between the age of ten years and the end of skeletal growth [13]. Patil, et al. reported that 51,911 patients underwent surgery to correct AIS in the United States from 1993 to 2002 [14]. At an average of $35,000 per case [15], the total surgical cost of AIS treatment in the United States is $182 million annually.

Scoliosis is diagnosed using the Adams Forward Bending Test. The person being screened is asked to stand with feet together and bend over from the waist leaving the arms dangling in front. The screener then measures the trunk rotation with a scoliometer (Figure 2.9). The amount of spinal curvature is measured using the Cobb method (Figure 2.9).

Figure 2.9 – (A) Adams forward bend test, (B) standing radiograph of a scoliotic spine (Reprinted with permission [37])
2.10). The angle is measured between lines drawn parallel the upper end plate of the superior end vertebra and the lower end plate of the inferior end vertebra of the curve of largest magnitude [16]. Curves having a Cobb angle greater than 10° are considered scoliotic, and surgery is indicated for extreme cases in which the Cobb angle exceeds 50° [17]. Surgery is warranted in such cases on the basis that 1) angles larger than 50° progress even after skeletal maturity, 2) larger curves can lead to loss of pulmonary function or respiratory failure, and 3) because surgery becomes more difficult as the curve magnitude increases [18].

Figure 2.10 – The Cobb angle
(Reprinted with permission [19])

2.8 Surgical Treatment of AIS

Among the treatments for AIS are a number of surgical approaches and techniques through which the lateral deviation is reduced and the anomalous rotation corrected (de-rotated). Deformity correction can be managed using either an anterior or posterior surgical approach or by a combination thereof. The combined approach will not be discussed as it is outside the scope of the current study.
2.8.1. Anterior Approach

The anterior surgical approach is well established in scoliosis correction. However, even with the introduction of video assisted thoracoscopic surgery, the use of the anterior technique has decreased due to risks to the thoracic aorta and other vital structures that are disturbed during an anterior approach [18]. This approach has been associated with pulmonary effusion, pneumothorax and embolus. Moreover, patients having surgical correction by the anterior approach have been known to have greater blood loss and surgery time as well as longer in-patient hospital stays [3]. When cases involve a significantly lordotic idiopathic curve or deficient posterior elements, however, an anterior approach is indicated [19].

2.8.2. Posterior Approach – Bilateral Medial Facetectomy

One posterior only technique used in deformity correction is the bilateral medial facetectomy (BMF) which involves bilateral resection of the inferior articular process as shown in Figure 2.11.

Figure 2.11 – Resection guidelines for bilateral medial facetectomy
The main advantage of this procedure is that it can be performed with minimal blood loss and without risk to neural elements. This method, however, produces only a small amount of correction per resected level, and it must be carried out at multiple levels [20].

2.8.3. Posterior Approach - Ponte Osteotomy

Another posterior-only approach to deformity correction was advocated by Alberto Ponte in the late 1980s in the treatment of Scheuermann’s Kyphosis. The Ponte osteotomy, which is depicted in Figure 2.12, involves a series of wedge segment osteotomies. Alberto Ponte described the procedure as creating a “compression across the osteotomy site posteriorly and height increase through the disk space anteriorly without complete disruption of the disk or anterior longitudinal ligament [21].” The procedure facilitates a posterior release that is beneficial in both coronal and sagittal plane realignment [22]. PO involves resection of the spinous processes and ligamentum flavum as well as both the superior and inferior articular facets at all surgical levels (Figure 2.13)
Removal of these ligaments was particularly important for axial rotation of the spine because the posterior elements are believed to provide the major resistance to this type of motion [24].

The Ponte procedure is performed on multiple levels as shown in Figure 2.14. Once the osteotomies are completed, the spinal column is stabilized with pedicle screw instrumentation (Figure 2.15).
Figure 2.14 – Multilevel Ponte osteotomies

Figure 2.15 – A schematic diagram of PO with stabilizing instrumentation (Reprinted with permission) [20]
The benefits of Ponte osteotomy include lower morbidity, shorter operating times and less blood loss than other approaches [3, 22, 25], and the main advantage of Ponte osteotomy over BMF is a larger range of motion, which allows for better curve correction.
3.1 Modification to Spine Flexibility-Testing Machine

The first section of this study involved an improvement to the counterweight mechanism of the spine machine.

It was desirable for pure moments to be applied consistently to the specimen. Therefore, an actively controlled counterweight system designed to reduce undesired shear loads was implemented as a means of moving a 2.5 kg mass with respect to a target. The target, the proximal end of the spinal segment, was expected to move within an area of 200 x 200 mm and at a speed of less than or equal to 5 mm per second.

A schematic illustration of the proposed positioning and counterweight system is shown in Figure 3.1. A motorized XY bi-slide (Model # XN10-0140-M02-71, Velmex, Inc.) was introduced to position the counterweight directly above the target. The bi-slide was mounted 1.5 meters directly above the specimen leaving adequate space to mount the torque arm vertically just above the specimen for testing in axial rotation. The XY bi-slide was aligned with the axes of the test machine.

The bi-slide servomotors (Vexta Type 17, 2 Phase, Single Shaft Stepper Motor, Velmex, Inc.) were selected based on several criteria. The travel distance required was 356 mm in X and in Y in order to accommodate the potentially wide range of motion of
an eight-segment specimen. Travel speed of up to 5 mm/s was needed to coincide with the specimen loading rate. It was also required that the slide be mounted upside-down and be able to support a 2.5kg mass that would be suspended from the bi-slide.

While this code successfully moved the bi-slide in the appropriate directions, for reasons not fully understood, the bi-slide would change directions indiscriminately. As there was not adequate time to resolve the issue, for this study the bi-slide motors were controlled manually.

3.2 Biomechanical Study
3.2.1. Study Design

Figure 3.1 – Schematic diagram of implemented active counterweight system

A repeated measures in vitro biomechanical flexibility-testing study on ten T11-L5 bovine specimens was performed using an existing spine flexibility-testing machine under the following sequential conditions: 1) intact spine, 2) after BMF, and 3) after
Ponte osteotomy. All surgical procedures were performed by the same faculty member of the Biomedical Engineering Department to avoid intra-operator variability. In addition, all surgical procedures were performed while the specimen was mounted to the test machine to avoid the need for recalibration of motion tracking equipment.

Each T11-L5 specimen was subjected to biomechanical testing:

1. While in its intact state (Figure 3.2A),

2. After facetectomy across T12-L1, L1-L2, L2-L3, L3-L4, and L4-L5 (Figure 3.2B), and

3. After Ponte osteotomy across T12-L1, L1-L2, L2-L3, L3-L4, and L4-L5 (Figure 3.2C).

Figure 3.2 – Specimen in each surgical state
(A) Intact specimen, (B) specimen after BMF, (C) specimen after Ponte osteotomy

3.2.2. Specimen Preparation

In preparation for the clinical study, ten 20-week-old bovine spine specimens that
had been stored in -20°C freezers [26] were dissected into T11-L5 segments and both end vertebræ potted in dental stone in preparation for bolting onto the test machine. The specimens were thawed at 3°C in a refrigerator for 12 hours followed by thawing at room temperature at about 20°C for an additional six hours prior to biomechanical tests.

After thawing, each specimen was dissected removing the muscle tissue from the vertebral column, and then potted in dental stone for bolting onto the test machine. Care was taken to create a small channel through the dental stone bilaterally and slightly anterior to the transverse processes for pass-through of the follower load cable. The proximal and distal ends of the specimen were anchored into the dental stone with wood screws and wire to prevent the specimen from pulling out of the hardened dental stone under testing conditions. The final step in specimen preparation was to insert eyelet screws bilaterally into each vertebral body from T12 to L5. The screws were placed in the lowest third of the vertebral body and inserted at a point slightly inferior and anterior to the pedicles with screw tips directed anteromedially. The eyelet screws served as guides for the follower load cables.

3.2.3. Flexibility Testing

The improved spine flexibility-testing machine as described in the previous section was used to execute biomechanical tests in three orthogonal directions on the multi-segmented spinal specimens. The specimens were tested cyclically in flexion-extension (FE) (with and without a 660N compressive follower load), lateral bending (LB) and axial rotation (AR) to address the aims of the clinical study.
The biomechanical tests conducted on each specimen in the three surgical states involved the application of loads to simulate physiological movements. The multi-segmented specimen was mounted on the testing machine by rigidly attaching the most distal vertebra to the testing machine platform. Before tightening the bolts, two laser levels were used to align the specimen properly. A pure moment was applied to T11 while L5 was rigidly fixed to the base of the testing machine. This moment applied in the appropriate directions caused the spine to undergo FE, LB or AR. Tests were carried out in these three orthogonal directions in random order. FE was tested twice, once with and once without a 660 N compressive follower load. A compressive pre-load of 660 N was selected because it is within the range of loads that simulate the physiologic load of the upper body on the human spine when standing upright [7, 10]. The test machine was operated at a rate of two degrees of rotation per second, thereby moving the spine in left and right lateral bending, left and right axial rotation, and flexion-extension for four full cycles of continuous motion with up to ±5 Nm of pure moment. These loading conditions were consistent with those typically used in biomechanical spine testing [7]. Previous biomechanical studies of human cadaveric cervical and lumber specimens have applied moments of ±1.5 and ±7.5 Nm, respectively [27, 1]. Since the current bovine specimens were long (seven vertebral levels), an applied moment of ±5 Nm was used so that sufficient force was applied to induce the physiological motions, but not enough force to damage the specimen. In order to minimize viscoelastic effects, the range of motion of the spine in each state was determined from the fourth test cycle. The ROM values for each specimen were normalized against their intact values. Each specimen thus acted as its own control thereby reducing the effects of inter-specimen variability.
3.2.4. Data Capture and Measurement

The Optotrak Certus optoelectronic camera system (Figure 3.3) was used to measure the three-dimensional (3D) kinematics of each vertebral body during tests. The 3D motion was measured via infrared emitting diodes (IREDs) rigidly attached to each vertebral body (T12, L1, L2, L3, L4, and L5), except the cranial vertebral level of the specimen, using Kirschner wires (K-wires) (Figure 3.4). Overall T12-L5 motion was analyzed. This system has an accuracy of ±0.1 mm and ±0.13° in 3D motion.

Figure 3.3 – Optotrak Certus high-speed camera
3.3 The Rigid Body

Each motion marker contained four IREDs encased in a Plexiglas base (Figure 3.4). One marker was considered a rigid body since each contained four IREDs and only three are necessary to constitute a rigid body [28]. Since the relative motion from one IRED to another on the same marker was fixed, each rigid body could have its own local coordinate system. These rigid bodies were defined through a software program (NDI 6D Architect, Northern Digital, Inc.) as part of pretesting setup. The local coordinate system made it possible for the software’s rigid body algorithm to determine what data would be pertinent, and therefore, included in the motion calculations.

The raw motion data were recorded in the built-in software package (NDI First Principles, Northern Digital, Inc.) used by the Optotrak Certus camera system. These data
were used by this program’s internal algorithms to quantify translational (T_x, T_y, T_z) and rotational (R_x, R_y, R_z) motions. The motions were calculated as one coordinate system (rigid body) with respect to other local coordinate systems (intersegmental ROM) and to a fixed coordinate system (overall ROM).

3.4 Data Analysis

In post-test analyses, overall ROM across T11-L5 in each surgical state was compared. The ROM was quantified as the distance between the maximum and minimum displacement during the fourth cycle. The percentage increase in ROM of T11-L5 in axial rotation following BMF represents the effects of facet joint release on the ability of the spine to be de-rotated. This result provides information on the degree of scoliosis correction that can be achieved with BMF. Similarly, the percentage increase in ROM of T11-L5 in AR, FE and LB following Ponte osteotomy was determined and compared against respective mean values. The results provide quantitative values on the effectiveness of further resection of the ligamentum flavum, interspinous ligament, inferior articular facet, and ligamentous facet capsule (Ponte osteotomy) at enabling increased deformity correction of the spine as compared to BMF. This result is important in determining whether or not the risks of Ponte osteotomy are justified by enabling increased ROM of the spine.

3.5 Statistical Analysis

A paired t-test (α = 0.05) was performed on the percent increase in ROM between BMF and Ponte osteotomy [7]. The statistical analysis was performed using SAS 9.2.
4.1 Overview

A repeated measures *in vitro* biomechanical flexibility-testing study on ten T11-L5 bovine specimens was performed under the following conditions: 1) intact spine, 2) after BMF, and 3) after Ponte osteotomy.

Mean T12-L5 ROM in AR, FEFL, FE and LB under ±5 Nm of applied moments were 8.4° (standard deviation = 2.7°), 17.7° (s.d. = 8.7°), 25.7° (s.d. = 8.1°) and 50.1° (s.d. = 12.3°), respectively, for the intact specimens. After BMF, corresponding mean ROM increased to 18.9° (s.d. = 6.6°), 20.0° (s.d. = 11.6°), 32.8° (s.d. = 10.0°) and 54.8° (s.d. = 12.3°), respectively. Further increases in mean ROM were observed after Ponte osteotomy, and the corresponding mean ROM were 32.4° (s.d. = 7.3°), 27.4° (s.d. = 16.1°), 45.9° (s.d. = 11.7°) and 56.9° (s.d. = 12.9°), respectively in AR, FE and LB. The study results were summarized in the chart in Figure 4.1. The load-displacement plots exhibited an “S-shaped” hysteresis curve, as expected. The fourth cycle curves for a single specimen in FEFL at the intact state, following facetectomy, and following Ponte osteotomy is shown in Figure 4.2.
Figure 4.1 – Overall change in ROM in all directions for each surgical state, standard errors indicated.

Figure 4.2 – Load displacement curve indicating increases in fourth cycle ROM. The largest increase in ROM resulted from PO when compared to the intact state. BMF also showed an increase in ROM over the intact state.
4.2 Overall ROM in Axial Rotation

The largest increase in range of motion was observed in axial rotation (Table 4.1). Compared to the intact results, there was a significantly greater increase in AR ROM after Ponte osteotomy (406% of intact) than after BMF (238% of intact) (p = 0.00009). Ponte osteotomy resulted in a mean increase of 23.9° (s.d. = 5.9°) in AR ROM across T12-L5 while BMF only resulted in a mean increase of 10.4° (s.d. = 6.1°). The results showed an increase of 110% in AR ROM after PO as compared to BMF.

Table 4.1 - Raw ROM Data (Degrees) for Axial Rotation

<table>
<thead>
<tr>
<th>Specimen ID</th>
<th>Intact T12-L5 (deg)</th>
<th>Fac. T12-L5 (deg)</th>
<th>Ponte T12-L5 (deg)</th>
<th>Fac vs Intact (%)</th>
<th>Ponte vs Intact (%)</th>
<th>Fac vs Intact (deg)</th>
<th>Ponte vs Intact (deg)</th>
</tr>
</thead>
<tbody>
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<td>3</td>
<td>8.7</td>
<td>9.8</td>
<td>24.1</td>
<td>112%</td>
<td>276%</td>
<td>1.1</td>
<td>15.3</td>
</tr>
<tr>
<td>9</td>
<td>4.3</td>
<td>12.5</td>
<td>25.3</td>
<td>290%</td>
<td>589%</td>
<td>8.2</td>
<td>21.0</td>
</tr>
<tr>
<td>8</td>
<td>10.1</td>
<td>17.1</td>
<td>40.4</td>
<td>170%</td>
<td>400%</td>
<td>7.0</td>
<td>30.3</td>
</tr>
<tr>
<td>6</td>
<td>7.4</td>
<td>15.3</td>
<td>26.8</td>
<td>207%</td>
<td>364%</td>
<td>7.9</td>
<td>19.5</td>
</tr>
<tr>
<td>5</td>
<td>10.3</td>
<td>19.4</td>
<td>35.4</td>
<td>188%</td>
<td>344%</td>
<td>9.1</td>
<td>25.1</td>
</tr>
<tr>
<td>1</td>
<td>6.2</td>
<td>16.5</td>
<td>35.2</td>
<td>265%</td>
<td>566%</td>
<td>10.3</td>
<td>29.0</td>
</tr>
<tr>
<td>2</td>
<td>12.8</td>
<td>19.2</td>
<td>35.1</td>
<td>150%</td>
<td>274%</td>
<td>6.4</td>
<td>22.3</td>
</tr>
<tr>
<td>4</td>
<td>7.7</td>
<td>22.0</td>
<td>30.5</td>
<td>287%</td>
<td>398%</td>
<td>14.4</td>
<td>22.8</td>
</tr>
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<td>7</td>
<td>11.4</td>
<td>33.5</td>
<td>46.1</td>
<td>294%</td>
<td>404%</td>
<td>22.1</td>
<td>34.7</td>
</tr>
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<td>23.5</td>
<td>25.0</td>
<td>418%</td>
<td>445%</td>
<td>17.9</td>
<td>19.4</td>
</tr>
<tr>
<td>Mean</td>
<td>8.4</td>
<td>18.9</td>
<td>32.4</td>
<td>238%</td>
<td>406%</td>
<td>10.4</td>
<td>23.9</td>
</tr>
</tbody>
</table>

| Std Dev     | 2.7                 | 6.6              | 7.3                | 90%               | 106%              | 6.1                | 5.9                |

4.3 Overall ROM in Flexion-Extension

For ROM in FE (Table 4.2), trials presented similar outcomes having a significantly larger increase in FE ROM after Ponte osteotomy (186% of intact) than after BMF (129 % of intact) (p = 0.001) alone. Ponte osteotomy resulted in a mean increase of 20.3 ° (s.d. = 8.7 °) in FE ROM while BMF resulted in a mean increase of only 7.1 ° (s.d. = 4.7 °).
Table 4.2 - Raw ROM Data (Degrees) for Flexion-Extension

<table>
<thead>
<tr>
<th>Specimen ID</th>
<th>Intact T12-L5 (deg)</th>
<th>Fac. T12-L5 (deg)</th>
<th>Ponte T12-L5 (deg)</th>
<th>Fac vs Intact (%)</th>
<th>Ponte vs Intact (%)</th>
<th>Fac vs Intact (deg)</th>
<th>Ponte vs Intact (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>15.2</td>
<td>15.0</td>
<td>28.3</td>
<td>99%</td>
<td>187%</td>
<td>-0.2</td>
<td>13.1</td>
</tr>
<tr>
<td>9</td>
<td>17.5</td>
<td>27.7</td>
<td>33.2</td>
<td>158%</td>
<td>189%</td>
<td>10.2</td>
<td>15.7</td>
</tr>
<tr>
<td>8</td>
<td>22.1</td>
<td>27.1</td>
<td>62.2</td>
<td>123%</td>
<td>282%</td>
<td>5.0</td>
<td>40.1</td>
</tr>
<tr>
<td>6</td>
<td>35.3</td>
<td>41.3</td>
<td>54.6</td>
<td>117%</td>
<td>155%</td>
<td>6.0</td>
<td>19.3</td>
</tr>
<tr>
<td>5</td>
<td>25.2</td>
<td>38.3</td>
<td>53.8</td>
<td>152%</td>
<td>213%</td>
<td>13.1</td>
<td>28.6</td>
</tr>
<tr>
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<td>17.2</td>
<td>25.9</td>
<td>38.7</td>
<td>151%</td>
<td>225%</td>
<td>8.8</td>
<td>21.5</td>
</tr>
<tr>
<td>2</td>
<td>30.1</td>
<td>31.9</td>
<td>47.4</td>
<td>106%</td>
<td>157%</td>
<td>1.7</td>
<td>17.3</td>
</tr>
<tr>
<td>4</td>
<td>30.3</td>
<td>33.5</td>
<td>40.3</td>
<td>111%</td>
<td>133%</td>
<td>3.2</td>
<td>10.0</td>
</tr>
<tr>
<td>7</td>
<td>39.3</td>
<td>52.0</td>
<td>61.1</td>
<td>132%</td>
<td>155%</td>
<td>12.6</td>
<td>21.8</td>
</tr>
<tr>
<td>10</td>
<td>24.6</td>
<td>35.7</td>
<td>39.9</td>
<td>145%</td>
<td>162%</td>
<td>11.0</td>
<td>15.3</td>
</tr>
<tr>
<td>Mean</td>
<td>25.7</td>
<td>32.8</td>
<td>45.9</td>
<td>129%</td>
<td>186%</td>
<td>7.1</td>
<td>20.3</td>
</tr>
<tr>
<td>Std Dev</td>
<td>8.1</td>
<td>10.0</td>
<td>11.7</td>
<td>21%</td>
<td>44%</td>
<td>4.7</td>
<td>8.7</td>
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</tbody>
</table>

4.4 Overall ROM in Flexion-Extension with Compressive Follower Load

The ROM in FEFL was smaller than in FE, as expected. The increase in ROM in FEFL was not as large as that shown in AR, but a significant increase was still observed (Table 4.3). FEFL trials showed a significantly larger increase in FEFL ROM after Ponte osteotomy (151% of intact) than after BMF (110% of intact) (p = 0.01) alone. Ponte osteotomy resulted in a mean increase of 9.7° (s.d. = 8.6°) in FEFL ROM while BMF resulted in a mean increase of only 2.3° (s.d. = 4.0°).
Table 4.3 - Raw ROM Data (Degrees) for Flexion-Extension With a 660 N Follower Load

<table>
<thead>
<tr>
<th>Specimen ID</th>
<th>Intact T12-L5 (deg)</th>
<th>Fac. T12-L5 (deg)</th>
<th>Ponte T12-L5 (deg)</th>
<th>Fac vs Intact (%)</th>
<th>Ponte vs Intact (%)</th>
<th>Fac vs Intact (deg)</th>
<th>Ponte vs Intact (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
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<td>NA</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
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<tr>
<td>9</td>
<td>11.7</td>
<td>15.1</td>
<td>26.9</td>
<td>129%</td>
<td>230%</td>
<td>3.4</td>
<td>15.2</td>
</tr>
<tr>
<td>8</td>
<td>12.5</td>
<td>13.2</td>
<td>26.8</td>
<td>106%</td>
<td>215%</td>
<td>0.7</td>
<td>14.3</td>
</tr>
<tr>
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<td>30.3</td>
<td>32.8</td>
<td>49.7</td>
<td>108%</td>
<td>164%</td>
<td>2.5</td>
<td>19.4</td>
</tr>
<tr>
<td>5</td>
<td>16.5</td>
<td>12.9</td>
<td>21.6</td>
<td>78%</td>
<td>131%</td>
<td>-3.6</td>
<td>5.1</td>
</tr>
<tr>
<td>1</td>
<td>9.9</td>
<td>9.8</td>
<td>11.0</td>
<td>99%</td>
<td>112%</td>
<td>-0.1</td>
<td>1.2</td>
</tr>
<tr>
<td>2</td>
<td>12.5</td>
<td>11.8</td>
<td>9.1</td>
<td>94%</td>
<td>72%</td>
<td>-0.7</td>
<td>-3.5</td>
</tr>
<tr>
<td>4</td>
<td>9.8</td>
<td>11.6</td>
<td>11.0</td>
<td>119%</td>
<td>113%</td>
<td>1.9</td>
<td>1.3</td>
</tr>
<tr>
<td>7</td>
<td>31.6</td>
<td>39.8</td>
<td>49.4</td>
<td>126%</td>
<td>156%</td>
<td>8.2</td>
<td>17.8</td>
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<td>136%</td>
<td>168%</td>
<td>8.6</td>
<td>16.5</td>
</tr>
<tr>
<td>Mean</td>
<td>17.7</td>
<td>20.0</td>
<td>27.4</td>
<td>110%</td>
<td>151%</td>
<td>2.3</td>
<td>9.7</td>
</tr>
<tr>
<td>Std Dev</td>
<td>8.7</td>
<td>11.6</td>
<td>16.1</td>
<td>18%</td>
<td>51%</td>
<td>4.0</td>
<td>8.6</td>
</tr>
</tbody>
</table>

4.5 Overall ROM in Lateral Bending

The results in LB were less marked but still indicated a significantly greater increase in ROM after Ponte osteotomy (117% of intact) than after BMF (113% of intact) (p = 0.049) (Table 4.4). Ponte osteotomy resulted in a mean increase of 7.4° (s.d. = 6.2°) in LB ROM while BMF resulted in a mean increase of only 5.3° (s.d. = 4.9°).
Table 4.4 - Raw ROM Data (Degrees) for Lateral Bending

<table>
<thead>
<tr>
<th>Specimen ID</th>
<th>Intact T12-L5 (deg)</th>
<th>Fac. T12-L5 (deg)</th>
<th>Ponte T12-L5 (deg)</th>
<th>Fac vs Intact (%)</th>
<th>Ponte vs Intact (%)</th>
<th>Fac vs Intact (deg)</th>
<th>Ponte vs Intact (deg)</th>
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<td>3</td>
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<td>42.8</td>
<td>42.4</td>
<td>124%</td>
<td>123%</td>
<td>8.3</td>
<td>7.8</td>
</tr>
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<td>24.6</td>
<td>33.9</td>
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<td>137%</td>
<td>9.3</td>
<td>9.1</td>
</tr>
<tr>
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<td>47.6</td>
<td>59.6</td>
<td>69.4</td>
<td>125%</td>
<td>146%</td>
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<td>62.4</td>
<td>66.0</td>
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<td>112%</td>
<td>3.6</td>
<td>7.2</td>
</tr>
<tr>
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<td>65.5</td>
<td>64.4</td>
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</tr>
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<td>43.7</td>
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</tr>
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<td>58.4</td>
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</tr>
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<td>56.7</td>
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<td>110%</td>
<td>2.6</td>
<td>5.0</td>
</tr>
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<td>69.9</td>
<td>75.9</td>
<td>76.2</td>
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<td>6.2</td>
</tr>
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<tr>
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<td>54.8</td>
<td>56.9</td>
<td>113%</td>
<td>117%</td>
<td>5.3</td>
<td>7.4</td>
</tr>
<tr>
<td>Std Dev</td>
<td>13.0</td>
<td>12.3</td>
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<td>14%</td>
<td>15%</td>
<td>4.9</td>
<td>6.2</td>
</tr>
</tbody>
</table>
CHAPTER V

DISCUSSION AND CONCLUSIONS

The purpose of this biomechanical study was to quantify and compare the range of motion realized by two surgical techniques used in the correction of adolescent idiopathic scoliosis. The current study found a significant increase in ROM in two of the three orthogonal test directions with Ponte osteotomy as compared to bilateral medial facetectomy.

5.1 Discussion

The current study quantified the increase in three-dimensional ROM achievable with Ponte osteotomy over BMF. This information is necessary for surgeons to be able to balance the pros and cons of a more extensive osteotomy technique against the increased magnitude in ROM achievable. As expected, a significantly higher ROM in AR and FE was found with PO than with BMF. The increase in LB ROM with Ponte osteotomy over facetectomy was not significant.

Similar to the current study, previous studies with sequential surgical release of posterior structures found increased flexibility of the spinal column [29, 30, 31, 32]. The results of the current study are in agreement with two previous studies that found significantly higher flexibility in FE and AR after removal of the posterior elements [33,
Increased flexibility with surgical releases when applied to scoliosis correction translates into larger deformity correction achievable through surgery.

Shufflebarger and Clark (1998) reported in a clinical study that Ponte osteotomy resulted in a 76% correction in Cobb angles in eight patients with AIS. In a follow-up study, Shufflebarger et al. (2004) reported Cobb angle reduction with Ponte osteotomy from 52° to 10° (80% correction) in a series of 62 patients with AIS. This translates into a mean of 6.0° of correction across seven motion segments or 8.4° across five motion segments. While absolute values obtained from young bovine spines in the current study cannot be directly interpreted as achievable degrees for the patient population, it is interesting to note that the current study quantified the achievable correction with Ponte osteotomy across each level in the young bovine spine as 6.5° in AR, 5.5° in FE and 11.3° in LB.

The three-dimensional flexibility of the intact young bovine specimens obtained in the current study is comparable with results from previous studies. Mean ROM across a single young bovine intact spinal motion segment in the current study was 1.7° in AR, 3.5° in FE and 9.9° in LB. Riley, et al., using an applied moment of ±6.4 Nm, measured mean ROM across L3-L4 of five 19-week-old bovine spines to be 2.1° in AR, 5.4° in FE and 10.4° in LB, and mean L3-L4 ROM across five elderly human spines (mean age = 68) to be 4.7° in AR, 6.3° in FE and 7.1° in LB [35]. Kettler, et al., using a higher applied moment of ±7.5 Nm, measured mean ROM across L1-L2 of six young bovine spines (9-12 months old) to be 2.1° in AR, 4.6° in FE and 10.5° in LB, and mean ROM across six elderly human spines (under 60 years old) to be 1.3° in AR, 5.9° in FE and 6.9° in LB.
These previous results indicated that bovine lumbar spines have a higher ROM in LB than human spines while ROM in FE and AR has similar ranges.

It is nearly impossible to obtain a reasonable sample size of human cadaveric specimens with similar age range (13 to 20 years old) and conditions (AIS) as the target scoliosis-correction patient population for in-vitro studies such as the current study. The available alternatives are to use either animal models or elderly human specimens. A previous study comparing flexibility of calf, pig, sheep and human spines found the young bovine spine to be the best animal model to predict effects of instrumentations in elderly human spines [36]. The effect of BMF and PO on spinal flexibility is assumed in the current study to have similar trends in the young thoracolumbar bovine spines and in the young thoracolumbar human spines with scoliosis. In agreement with this assumption, a previous study found that facetectomy resulted in an increase of 12% in FE ROM in elderly human cadaveric specimens [32] while the current study using young bovine spines measured an increase of 10% in FE ROM.

5.2 Assumptions and Limitations

The results of this study showed that, while BMF yielded an increase in ROM in all three directions over the intact specimen, Ponte osteotomy provided an even greater increase in ROM. Some limitations of this study necessitate future work. Some suggestions for future studies are as follows:

1. This bovine study should be followed by a human cadaveric study, as the absolute magnitude in deformity correction achievable in patients remains unclear.
2. Although reliable data were obtained from this study, future studies should be performed using thoracic specimens since AIS is seen mainly in the thoracic region and thoracolumbar regions.

3. The XY bi-slide used to position the counterweight on the test machine was operated manually in this study. Control of the bi-slide should be incorporated into the existing LabVIEW program used to control the spine-testing machine.

4. In further studies, a means of measuring the angle in the counterweight ropes should be incorporated so as not to rely on visual approximation of vertical alignment.

5.3 Study of Hypotheses

The results of this study showed a larger increase in ROM in each of the four loading conditions after Ponte osteotomy than after BMF. In the AR, FE, and FEFL conditions, the ROM after PO was significantly different from that of BMF. For this reason, null hypotheses (1), (3) and (4) are rejected and alternative hypotheses (1), (3), and (4) accepted. In the case of lateral bending, null hypothesis (2) is accepted, and alternative hypothesis (2) is rejected. Post-priori calculations showed that the power of the current study was greater than 95% for AR, FE and FEFL, and 52% for LB.

5.4 Conclusions

The ROM values achievable with BMF and PO were quantified and compared in the current study using a bovine model. Ponte osteotomy provided significant increases in
the achievable ROM in the flexion-extension (with and without follower load) and axial rotation bending planes as compared to bilateral medial facetectomy. Increases in ROM achieved through Ponte osteotomy were not significantly larger than those achieved through facetectomy.
REFERENCES


[27] N. Kumar, "In-vitro biomechanical analysis of Open Door Laminoplasty with partial or total posterior ligament resection," Akron, OH, 2011.


APPENDIX A

SAS STATISTICAL CODE & OUTPUT

/* Thesis AR(paired) */
data;
input facAR ponteAR;
/* The following line is for paired t-test */
diffAR=ponteAR-facAR;
datalines;
112 276
290 589
170 400
207 364
188 344
265 566
150 274
287 398
294 404
418 445
;
proc means mean stddev stderr t probt;
var diffAR;
run;
/* Thesis FE(paired) */
data;
input facFE ponteFE;
/* The following line is for paired t-test */
diffFE=ponteFE-facFE;
datalines;
99 187
158 189
123 282
117 155
152 213
151 225
106 157
111 133
132 155
145 162
;
proc means mean stddev stderr t probt;
var diffFE;
run;
/* Thesis FEFL(paired) */
data;
input facFEFL ponteFEFL;
/* The following line is for paired t-test */
diffFEFL=ponteFEFL-facFEFL;
datalines;
129 230
106 215
108 164
78 131
99 112
94 72
119 113
126 156
136 168
;
proc means mean stddev stderr t probt;
var diffFEFL;
run;
/* Thesis LB(paired) */
data;
input facLB ponteLB;
/* The following line is for paired t-test */
diffLB=ponteLB-facLB;
datalines;
124 123
138 137
125 146
106 112
123 121
100 112
99 98
105 110

```sas
proc means mean stderr t probt;
var diffLB;
run;
```

### Analysis Variable: `diffAR`

| Mean         | Std Dev       | Std Error    | t Value | Pr > |t| |
|--------------|---------------|--------------|---------|------|----------|
| 167.9000000  | 86.5415122    | 27.3668291   | 6.14    | 0.0002 |

### Analysis Variable: `diffFE`

| Mean         | Std Dev       | Std Error    | t Value | Pr > |t| |
|--------------|---------------|--------------|---------|------|----------|
| 56.4000000   | 43.1024619    | 13.6301952   | 4.14    | 0.0025 |

### Analysis Variable: `diffFEFL`

| Mean         | Std Dev       | Std Error    | t Value | Pr > |t| |
|--------------|---------------|--------------|---------|------|----------|
| 40.6666667   | 44.3790491    | 14.7930164   | 2.75    | 0.0251 |

### Analysis Variable: `diffLB`

| Mean         | Std Dev       | Std Error    | t Value | Pr > |t| |
|--------------|---------------|--------------|---------|------|----------|
| 4.3000000    | 7.2884993     | 2.3048259    | 1.87    | 0.0950 *|

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APPENDIX B

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