University of Alberta

The Effect of Torso Muscle Contraction on Lumbar Spinal Stability

by

Natasha Rachel Liddle ©

A thesis submitted to the Faculty of Graduate Studies and Research
in partial fulfillment of the requirements for the degree of

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in
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Dedication

To my best friend and husband, Chris

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Abstract:

The Effect of Torso Muscle Contraction on Lumbar Spinal Stability

Introduction: The purpose of this study was to determine the spinal stiffening effect of two different exercises frequently prescribed for low back pain.

Methods: Twenty-eight asymptomatic subjects were taught abdominal hollow and abdominal brace stability exercises. During periods of rest and contraction, stiffness of the lumbar spine was quantified in each subject using an assisted indentation technique. Additionally, electromyography and B-mode ultrasound were used to characterize trunk muscle activity.

Results: Spine stiffness was significantly greater for each contraction compared to rest ($p < 0.001$) while the abdominal brace generated significantly greater global and mean maximal stiffness compared to the abdominal hollow ($p = 0.022$, $p = 0.013$; respectively). No gender difference was noted.

Conclusion: The abdominal brace provided a greater stiffening effect to the spine. This finding may provide knowledge to better match exercises to specific patient needs.

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List of Symbols/Abbreviations

A/D	Analog to Digital
ANOVA	Analysis of Variance
AI	Assisted Indentation
BMI	Body Mass Index
cm	centimeters
cm^2	centimeters squared
CSA	Cross-sectional Area
DC	Direct Current
EMG	Electromyography
ES	Erector Spinae
EO	External Obliques
GS	Global Stiffness
Hz	Hertz
ICC	Intra-class Correlation Coefficient
IO	Internal Obliques
ITI	Inter-trial Inconsistency
LVDT	Linear Velocity Displacement Transducer
L-ES	Lumbar Erector Spinae
L3	Lumbar Vertebra Three
L4	Lumbar Vertebra Four
L5	Lumbar Vertebra Five
MRI	Magnetic Resonance Imaging
MVC	Maximum Voluntary Contraction
MMS	Mean Maximal Stiffness
mHz	millihertz
mm	millimeters
ms	milliseconds
N	Newtons

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Nm/rad	Newtons * meters/radians
N/mm	Newtons per millimeter
PA	Posteroanterior
RA	Rectus Abdominis
SPS	Spinal Physiotherapy Simulator
SPAM	Spinal Posteroanterior Mobilizer
SD	Standard Deviation
SAM	Stiffness Assessment Machine
sEMG	Surface Electromyography
T-ES	Thoracic Erector Spinae
TLF	Thoracolumbar Fascia
TrA	Transversus Abdominis
V	Volts

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CHAPTER ONE
INTRODUCTION

1.1 Overview:

Low back pain is a significant problem for which few causes are known. Given this lack of knowledge, numerous explanations for the cause of low back pain exist, including instability between spinal vertebrae. As a result, health care professionals frequently prescribe stabilization exercises to "improve" spinal stability. Unfortunately, the effectiveness of these exercises is unknown. This thesis has been designed to determine the effect of two clinically distinct exercises on spinal stability. Knowledge from this work will help guide clinical prescription of exercise and advance our understanding of spinal function.

1.2 Background and rationale:

It is estimated that 45% of the adult population experiences low back pain (LBP) each year[1] and that back pain recurs in an overwhelming 60-80% of patients within a year.[1-3] Given the morbidity associated with this prevalent condition, LBP is now ranked the second most common cause of disability in the United States[1] costing over $90.7 billion USD in 1998 alone.[4]

The high morbidity and cost of LBP are directly related to the lack of an effective treatment for the condition. Arguably, this deficiency in effective treatment is due in part to incomplete understanding of the cause of LBP – only 15% of LBP can be attributed to a specific etiology.[5] If the cause of back pain can not be ascertained, effective treatment specific to the cause cannot be provided, resulting in a high morbidity and an increased expenditure rate as numerous treatment options are trialed.

Because the causes of LBP are not yet fully defined, there is no shortage of suggestions for its etiology. While too numerous to list, several suggested causes of LBP have focused on the separate anatomical elements of the back (e.g. bone, muscle, nerve) and more importantly, how they work together (or fail to work together). For example, it has

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been shown that the ligamentous lumbar spine can withstand only 88N of vertical force before buckling, making it is an inherently unstable system requiring external support.[6] Because everyday activities present spinal loads in the realm of 1,000 – 3,400 N[7, 8] and 6,000-18,000 N for more extreme activities[9, 10] it is apparent that the majority of the stability of the spine comes from other systems such as muscular contraction and increases in intra-abdominal pressure. As the muscular system is the most active and adaptable of these systems, spinal stability created by muscular contraction has been a recent focus in attempting to understand low back pain and guide its treatment.

1.3 Spinal stability:

In the absence of mechanical damage to the spine, several theories have been advanced to explain how stability is achieved, the most prominent from Panjabi.[11] Panjabi theorized that stability of the spine is dependent upon contributions from the passive structures (ligaments and bone), active structures (muscle and tendon) and neural control of these structures (nerves and central nervous system). Thus, deficiencies in merely one or a combination of these subsystems could be responsible for the creation of instability and pain in the lumbar spine.

Although most would agree with the theory of Panjabi, there is disagreement as to exactly how muscles are used to achieve meaningful stability. Indeed, a review of the literature demonstrates that two major theories exist with respect to how muscles are used to generate and maintain spinal stability: local muscle contraction and global/co-contraction of numerous muscles. In the local muscle contraction theory, it is thought that deep abdominal and low back muscles are activated to provide an "abdominal corset" around the trunk that provides stability to the spine.[12-21] Conversely, the global/co-contraction theory argues that the co-ordinated contraction of *all* spinal muscles is required to provide stability, just as all guy-wires need to be connected to effectively hold a ship's mast in place.[22-27]

While on one level, these schools of thought appear to be similar, on a clinical level, the idea of local or global theory of instability has resulted in two different exercise

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prescriptions. These exercises are termed “abdominal hollowing” (local contraction muscle theory) and the “abdominal brace” (global/co-contraction muscle theory).

1.4 Deficiencies in the literature:

Numerous studies have been completed that implicate certain muscles or muscular patterns of contraction in contributing to spinal stability. However, these studies are limited in that there are very few methods available to quantify the effect of these contractions on spinal stability itself.

Specifically, several studies have demonstrated that a deep abdominal muscle (transverses abdominis or TrA) contracts prior to arm or leg movement.[17, 18] These studies conclude that these contractions act to stabilize the spine and create a stable platform for distal movements of the extremities.[17, 18] However, the effect of contraction by these muscles on spinal stability has not been measured directly; rather it has been assumed, based on the relative timing of a muscular contraction. Other studies have employed mathematical approaches to calculate the contribution of muscles to stabilization of the spine.[22, 23, 26, 28, 29] In this approach, muscular action on the spine is represented as a line of force and subsequent calculations of stability are made through comparisons of all forces acting on the spine.[30] Inherent limitations to this approach include the presence of inaccurate anatomical descriptions[31,32] as well as significant variation between individuals regarding the origin and insertion of torso muscles.[31, 32] Therefore, the hypothesized line of force of a muscle on a vertebral segment may not be accurate within each individual patient. Finally, other studies have drawn an indirect relation between treatment success or patient satisfaction with respect to spinal stability.[20] In these studies, it is assumed that if patients report improvement with a prescribed stabilization program, their stability must have increased. However, stability itself has not been quantified in these studies, it is merely assumed to increase.

Given the above, there is a need to measure spinal stability in response to muscle contractions occurring directly in human subjects. Because it is unlikely that any single outcome measure could summarize the complexities of spinal stability, an alternative

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approach would be to measure discrete aspects of stability directly. With this in mind, several current techniques exist which could be employed to evaluate specific aspects of stability. Such is the case with a technique termed assisted indentation (AI). Using this method, a posteroanterior force is applied to the spine and the stiffness of the underlying tissues (soft tissue and bone) is calculated. Although not a complete measure of spinal stability, spinal stiffness does allow for the quantification of force required to disrupt equilibrium in the spine, thus quantifying an important aspect of stability.

1.5 Statement of the problem

Due to the costly nature of low back pain, it becomes paramount that the interventions used to treat this condition have scientific evidence for their effectiveness. Currently, numerous clinical interventions are prescribed to patients on the premise of improving spinal stability. However, very few human studies have been completed that use an objective physical measure of spinal stiffness to determine if muscular contractions alter spinal stability. Therefore, in order to establish the scientific merit behind these interventions, stability in the spine must be quantified and the effect of these clinical interventions on stability determined.

1.6 Objectives:

The objectives of this thesis are as follows:

1) To verify the reliability of measurements used in this project (e.g. AI and ultrasound measurements).
2) To quantify the change in spinal stiffness achieved by performance of two stabilizing torso contractions, (abdominal hollowing and abdominal bracing) in a healthy, adult population.

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1.7 Hypotheses:

The research hypotheses of this thesis are based on a review of the literature as well as the author's clinical experience. The hypotheses are as follows:

1) The reliability of measures used to assess spinal stability will be excellent (ICC greater than or equal to 0.75).
2) The reliability of measures used to assess muscle contraction (TrA) will be excellent (ICC greater than or equal to 0.75).
3) Two exercises used clinically (hollow and brace) will increase the stiffness of the lumbar spine significantly in a healthy, adult population, as measured by AI.
4) Abdominal hollowing will increase spinal stiffness to a greater extent than abdominal bracing in a healthy, adult population, as measured by AI.

1.8 Definitions:

Abdominal brace – An isometric contraction of all the muscles of the abdominal wall and low back without a change in the position of the muscles or trunk.[26, 33]

Abdominal hollowing – A selective recruitment of the transverse abdominis while minimizing activation of the rectus abdominis and the obliques.[34] This stabilizing exercise also involves a co-contraction of the lumbar multifidus.[34]

Assisted indentation – A procedure used to determine stiffness of the spine and soft tissues that mimics digital compression used in manual palpation. A posteroanterior force is applied to the spine using an indentor. Force and displacement properties are quantified during this process[35] using a load cell and a linear variable displacement transducer (LVDT), respectively.

Active control system – consists of the muscles and tendons surrounding and acting on the spinal column.[11]

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B-mode ultrasound – refers to brightness-mode ultrasound where the brightness of the image corresponds to the amplitude or strength of the reflected signal. In B-mode ultrasound, multiple A-mode channels are used to produce a two-dimensional graphical display of the underlying tissue. The distance between the areas of brightness on the graphical display corresponds to the distance between the reflecting interfaces of the tissues being imaged.[36]

Creep – Characteristic of viscoelastic materials, such that when a load is suddenly applied and then kept constant thereafter; increased deformation of the viscoelastic material occurs over time.[37] The deformation-time curve approaches a steady-state value asymptotically.[37]

Concurrent validity – Controversial definitions exist regarding this type of validity, ranging from the correlation of two measures taken at the same period of time to the correlation of a measure with a gold standard. For the purposes of this thesis, concurrent validity is defined as the degree to which the outcomes of a new measurement technique correlate to outcomes of a previously established measurement technique, such that significant differences should not be present between the two measurements.

Dynamic stability – For this thesis, dynamic stability will refer to the coordination of the passive, active, and neural control subsystems to maintain control over the neural zone of the spinal segments while allowing active movement.

Functional stability – In this thesis, functional stability will refer to the ability to maintain control over the neutral zone such that there are no limitations present regarding performance of activities of daily living.

Global muscles – the large, more superficial, multi-segmental muscles of the trunk that are primarily involved in generating spinal movement. Global muscles include: internal obliques, external obliques, rectus abdominis, quadratus lumborum (lateral fibers), iliocostalis lumborum pars thoracis, and longissimus thoracis pars thoracis.[21, 34]

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Hysteresis – the loss of energy in the form of heat during each loading and unloading cycle of viscoelastic materials.[37]

Instability – a significant decrease in the capacity of the stabilizing systems of the spine to maintain intervertebral neutral zones within physiological limits so there is no major deformity, neurological deficit or incapacitating pain.[11, 38] Also termed clinical instability or translational instability. When discussing dynamic stability, instability can be defined as an inability to maintain a posture or control a movement.

Local muscles - the inter-segmental, deep muscles of the trunk that primarily work to control the stiffness and intervertebral relationships of the spinal segments. These muscles include the intertransversarii, interspinales, multifidus, quadratus lumborum (medial fibers), transversus abdominis, internal obliques (fiber insertion into thoracolumbar fascia), longissimus thoracis pars lumborum, and iliocostalis lumborum pars lumborum.[21, 34].

Neutral posture of the spine - The posture of the spine in which the overall internal stresses in the spinal column and the muscular effort to hold the posture are minimal.[38] Muscular contraction is thought to be especially important around the neutral posture during movement or load carrying situations of the spine[23] as the ligaments of the spine are not taut in this position thus requiring muscular support.

Neutral Zone – That part of the range of physiological intervertebral motion, measured from the neutral position, within which the spinal motion is produced with a minimal internal resistance. It is the zone of high flexibility or laxity.[38] Size of the neutral zone is influenced by the interaction between the passive, active and neural control systems.

Neural control system – comprised of the various force and motion transducers, located in ligaments, tendon, and muscles and the neutral control centers (nerves and central

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nervous system which direct and control the active system in providing dynamic stability).[11]

Passive control system – includes vertebrae, facet articulations, intervertebral discs, spinal ligaments, and joint capsules, as well as the passive mechanical properties of the muscles.[11]

Posteroanterior (PA) pressure test – A manual technique used to assess spinal stiffness. This technique involves manual application of an anteriorly directed force to the spinous process or transverse process of the vertebrae of a prone subject.[39]

Segment – A functional vertebral unit in the spine. A vertebral segment consists of two adjacent vertebrae joined by apophyseal joints and the intervertebral disc.

Spinal stability – For this thesis, spinal stability will refer to the coordination of the passive, active, and neural control systems to ensure proper control over the intervertebral neutral zone. It consists of dynamic, static, and functional stability.

Spinal stiffness – For this thesis, spinal stiffness will refer to the resultant vertebral displacement and soft tissue compression and stretching that occurs during a specified force application (100 N) at a set rate (2 mm/sec) over the spinous process of the spine. Stiffness will be quantified using AI and will represent a measurement of the stiffness of the 4th lumbar vertebrae due to surrounding tissue. In regards to a general definition, spinal stiffness involves the resistance of the spine (passive, active, and neuromuscular subsystems) to movement in a posteroanterior direction. It refers to the load-displacement relationship of the spine.[35] In this thesis, sub-types of stiffness will also be utilized. These can be stratified into 5 major categories based on the anatomy being tested, its location, tissue response, the intended use of the results, or the quantitative measurement of it. A further description of each sub-type at its parent category follows.

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- Anatomical:
  - *Spinal stiffness* – A general measure of stiffness in which the contribution of different components (e.g. soft tissue compression versus boney displacement) cannot be determined.
  - *Vertebral stiffness* – The relative amount of displacement of the vertebrae, not including compression of superficial soft tissue, that occurs with application of a specified force. This stiffness remains an approximation as tissue compression deep to the spine can occur with stiffness testing using manual indentation.
  - *Soft tissue stiffness* – The relative amount of compression of the superficial soft tissue that occurs with the application of a specified force. Soft tissues that have higher levels of stiffness will exhibit less compression with application of a specified force.
- Location:
  - *Regional stiffness* - A term used to describe the general stiffness of a region (e.g. lumbar spine).
  - *Segmental stiffness* - The stiffness value between two adjacent vertebrae in the spine. In this case, the displacement of the vertebral level of interest and the adjacent vertebral level are both quantified resulting in a measurement that reflects the stiffness between two specific levels. This stiffness includes the resistance to force of the adjacent two vertebrae (and the inert structures attached to these areas), the soft tissues of the spine, and the intervertebral disc.
- Tissue Response:
  - *Elastic stiffness* – Evident in material that when deformed by a load, returns to the original value without any loss of energy. For example, a spring exhibits elastic stiffness.[37, 40]
  - *Viscous stiffness* – Evident in material that does not return any energy stored during loading, rather energy is dissipated as heat. For example, a dashpot exhibits viscous stiffness.[37, 40]

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- Pragmatic:
  - *Clinical stiffness* - Stiffness of a vertebral level as identified and/or assessed by manual posteroanterior pressures on the spine. The clinical concept of spinal stiffness when using a manual posteroanterior pressure test to assess stiffness has been found to include two separate aspects: 1) Mobility and 2) Nature of resistance to movement.[41] Mobility is commonly delineated into: 1) normal; 2) hypomobile; 3) hypermobile.[42]
- Measurement:
  - *Global Stiffness (GS)* – The slope of the force-displacement curve between 30 N and maximal force (N/mm). It has been shown that the force-displacement properties of the spine, after 30 N of force, exhibit a linear relationship.[43-45] Also termed coefficient K in the literature.
  - *Mean Maximal Stiffness (MMS)* – The average of the stiffness values (N/mm) during the time period of the indentation when maximal force has been imparted to the vertebrae and held for a period of approximately 1 second. The stiffness acts as a ratio between the force imparted and the boney displacement of the spinous process at the same time period.
  - *Short-range Stiffness* – The stiffness during the initial force application of an indentation which reflects the stiffness of the passive structures (and the passive component of muscles through stretching of crossbridges). After a certain force (30 N), the reflex system of the muscular tissue surrounding the passive structure activates and the short-range stiffness is no longer being measured. This region of the force-displacement curve is non-linear in nature. Short-range stiffness is often measured as the displacement of the indentor from 0.5 N – 30 N. [43-45] Changes in the slope characteristics of this region of the force displacement curve may describe changes in passive tissue properties. Also termed D30 in the literature.

Stabilizing torso muscle contractions – Low-grade muscle contractions of the torso muscles (approximately 30% MVC) that are taught to patients to help improve spinal

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stability. For the duration of this paper, this term will include the contractions of the abdominal brace and abdominal hollowing.

Static stability – In this paper, static stability will refer to the coordination of the three subsystems to maintain control over the neutral zone when no spinal movement is occurring.

Stress – the force per unit area of a structure.[37]

Stress relaxation – Occurs in viscoelastic materials; the decrease in stress in a deformed structure with time when the deformation is held constant.[37]

Viscoelasticity – The time-dependent property of a material (e.g. hysteresis, creep, stress relaxation) that shows sensitivity to rate of loading or deformation.[37] Two basic components of viscoelasticity include viscosity and elasticity with the behavior of a viscoelastic material acting as a combination of these two properties. Two other practical phenomena are typical of viscoelastic materials: First, the load-deformation curve of a viscoelastic material is dependent upon the rate of loading. When higher rates of loading occur, it results in a steeper load-deformation curve. Second, viscoelastic materials exhibit loss of energy in the form of heat during each loading and unloading cycle (hysteresis).[37]

1.9 Limitations:

This study is limited by the use of AI to measure stiffness in the spine. Due to the difficulty in quantifying spinal stability, certain aspects of stability that change with abdominal muscle contraction patterns may not be measured by this tool, such as rotational instabilities. As well, the study does not utilize a random sample. Because a convenience sample will be used instead, it is possible that the study population will not be representative of the general population; therefore, measurement of subject characteristics is necessary in order to define the population to which the results of this study can be applied.

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The current study is valid only in the following conditions:

1. For healthy subjects without any current low back pain.
2. For subjects between the ages of 18-30.

1.10 Delimitations:

This study is delimited to:

1. Test subjects able to perform both abdominal patterns of contraction.

1.11 Ethical considerations:

There are two physical interventions that have risk potential in this study. The first intervention is AI. During indentation, subjects may experience some soreness in the low back from the indentation process. If soreness does occur, the circumstances surrounding the onset of soreness will be altered if possible (e.g. re-positioning of the indentor head). If this is not satisfactory, the subjects will be instructed to rest in a comfortable position until the soreness has resolved and will be instructed to ice the area. Soreness of the low back may come on right away or a may develop a few hours following stiffness testing; however, the care instructions to the subject will remain the same. Furthermore, the subject will be instructed to contact the researcher if soreness does occur at a later time. Indentation forces will not exceed 100 N, a force level commonly used in the literature and considered to be safe in the human lumbar spine.[44] Second, there is a very small chance that surface electromyography electrodes applied in this study will cause skin irritation. If this does occur, the subject will be instructed that this is a self-limiting problem that will resolve in approximately 1 week.

Informed consent was obtained from the subjects prior to enrollment in this study. Approval from the University of Alberta Health Research Ethics Board was obtained prior to the thesis' commencement. (See Appendix A).

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1.12 Significance of the study:

By further understanding the effect of stabilization exercises on the mechanics of the low back in an asymptomatic population, increased knowledge will be obtained regarding the mechanisms by which stabilization therapy may help patients with low back pain. It is possible that stabilization exercises work primarily by increasing the stiffness of the lumbar spine, thus providing a stabilizing force to the vertebral column during movements and reducing pain. However, the use of stabilization exercises may chiefly act as a reminder to patients to maintain proper posture during activities, as a result reducing overall stresses on the spine and decreasing pain. Investigation of this topic may provide clinicians and researchers with increased knowledge of the nature of spinal mechanics that should be present with stabilization exercises. This may allow for use of new outcome measures, such as stiffness assessment, to be used in concert with other clinical measures (such as assessment of pain, range of motion, strength) to get a better global picture of patient variables and the response of these variables to treatment. Further, the measurement of spinal stiffness could also act as a feedback mechanism for subjects. For example, perhaps the stiffness of the back should increase by a certain amount when the stabilization exercise is performed properly. This may also increase the possibility of implementing spinal stiffness measurement devices into the clinical environment as an adjunct to clinical experience.

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CHAPTER TWO
LITERATURE REVIEW

2.1 Anatomy and function of trunk musculature

As the model of spinal stability has evolved from one in which the osteoligamentous structures were viewed as primary stabilizers to one in which muscular control is seen as crucial, more emphasis has been placed on determining the specific role of muscles in providing stability. Due to the large number of muscles attaching both directly and indirectly to the lumbar spine, stability may in fact be achieved in many ways through muscular contraction. Functionally, numerous muscles, including TrA, psoas, quadratus lumborum, and lumbar multifidus have been described as contributing to the control of lumbar segmental motion either through development of intersegmental stiffness or through maintenance of spinal equilibrium.[15, 46, 47] Furthermore, several other muscles such as the erector spinae, abdominal obliques and rectus abdominis have been implicated in the contribution to dynamic stability through control of body position and equilibrium.[48] Therefore, the anatomical considerations and functions of these muscles is important when determining their contribution toward stability. For the purposes of this study, the focus will be placed on muscles that are thought to be recruited during the two stabilizing torso contractions (hollow and brace): transversus abdominis (TrA), multifidus, rectus abdominis (RA), external oblique (OE), internal oblique (OI), and the erector spinae (ES).

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Transversus abdominis

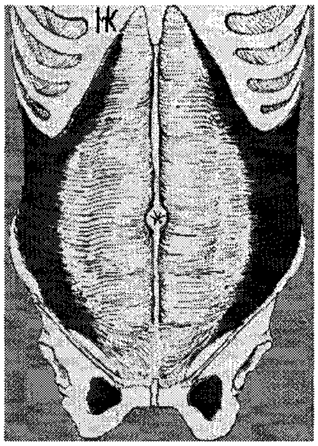

Figure 1: Anterior view of TrA.[49]

A deep abdominal muscle, TrA originates from the internal surfaces of the 7-12th costal cartilages, the middle and posterior layer of the thoracolumbar fascia, [50] iliac crest, and inguinal ligament, and inserts into the linea alba and public crest.[51] Due to this orientation, its primary function has been reported to aid in compression of the abdominal viscera.[51] It has been theorized that TrA may influence spinal stability by increasing intra-abdominal pressure (IAP) or tensioning the thoracolumbar fascia which inserts into the lumbar vertebrae.[15, 52] Given these functions, it is thought that upon contraction, this muscle acts as an abdominal corset that works to combat instability during dynamic movements.[34]

It has also been reported that the insertion location of TrA is variable. Numerous cases have been documented in which complete or partial detachment of this muscle from the iliac crest and/or absence of fascicles below the iliac crest were present.[31] This suggests that contraction of TrA may not perform the same function in different people. Furthermore, recent research has suggested that the primary assumption that TrA fibers are oriented in only a horizontal direction may be inaccurate. Evidence from a recent cadaveric dissection study by Urquhart et al.[31] demonstrated that the fascicle orientation of the upper region of TrA was horizontal, but that the middle and lower fascicles of TrA were inferomedially directed, implying a differentiation of function within the same

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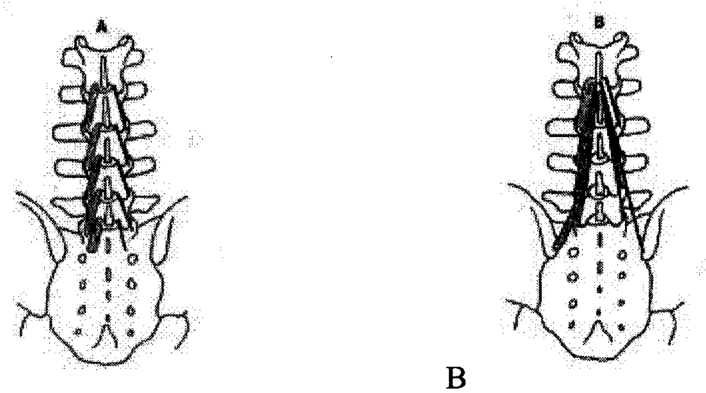
muscle. It is hypothesized that the upper fibers of the TrA work primarily on the rib cage, the middle fibers on the lumbar spine, and the lower fibers on the sacroiliac joint.

Multifidus:



A. B

Figure 2 : Posterior view of the deep Multifidus fibers (A) and of the superficial Multifidus fibers (B).[53]

Multifidus has been found to consist of five bands, each of which is made up of overlapping fascicles (deep and superficial) that radiate from the lumbar spinous processes and laminae to insert into the sacrum, posterior superior iliac spine (PSIS), and laminae of the caudal vertebrae.[54, 55] This muscle has been observed to span from the L1-S4 spinal levels.[54] The primary role of the multifidus appears to be to act directly on the lumbar vertebral column and produce the anti-flexion (extension) moment that is needed to counteract the anterior sagittal rotation generated by the contraction of the abdominal obliques.[51, 54, 55] Although originally thought to have a role in rotation and lateral flexion, it is suggested that multifidus plays a very insignificant role in these movements as both its superficial and deep fibers attaches too close to the axis of these movements to significantly contribute.[55] Further studies suggest that multifidus plays a proprioceptive role in the spine to protect the articular structures, discs, and ligaments from excessive bending strains and injury.[56]

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Rectus abdominis:

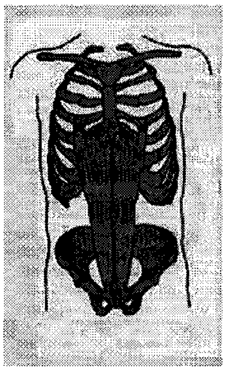

Figure 3: Anterior view of Rectus Abdominis.[57]

The central abdominal muscle, rectus abdominis, originates from the pubic symphysis and pubic crest to insert into the xiphoid process and 5-7th costal cartilages.[51] It primarily functions as a trunk flexor but also works to compress the abdominal viscera.[51]

Internal obliques:

Figure 4:Anterior view of the right Internal Oblique.[58]

The internal obliques (IO), originate from the thoracolumbar fascia, iliac crest, and lateral half of the inguinal ligament then insert on the inferior border of the 10-12th ribs, linea alba, and the pubis via the conjoint tendon.[51] Recent findings have shown that the fibers of the internal obliques run superomedially above the level of the iliac crest, and below the iliac crest the fibers begin to run horizontally with increasing inferomedial

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angulation.[31] The IO has also been shown to become two distinct muscle layers in the middle and lower regions of the abdomen.[31] Due to the increased inferomedial fascicle orientation and divisions in the lower muscle layers, it is suggested that in addition to the primary action of ipsilateral trunk rotation and flexion[51], the IO may also assist in compression of the sacroiliac joints.[31, 59]

External obliques:

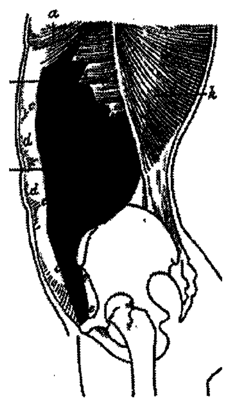

Figure 5: Lateral view of the left External Oblique.[60]

Assisting in trunk flexion and contralateral rotation, the external obliques originate from the external surfaces of the 5-12th ribs and insert onto the linea alba, pubic tubercle and anterior half of the iliac crest.[51]

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Erector spinae:

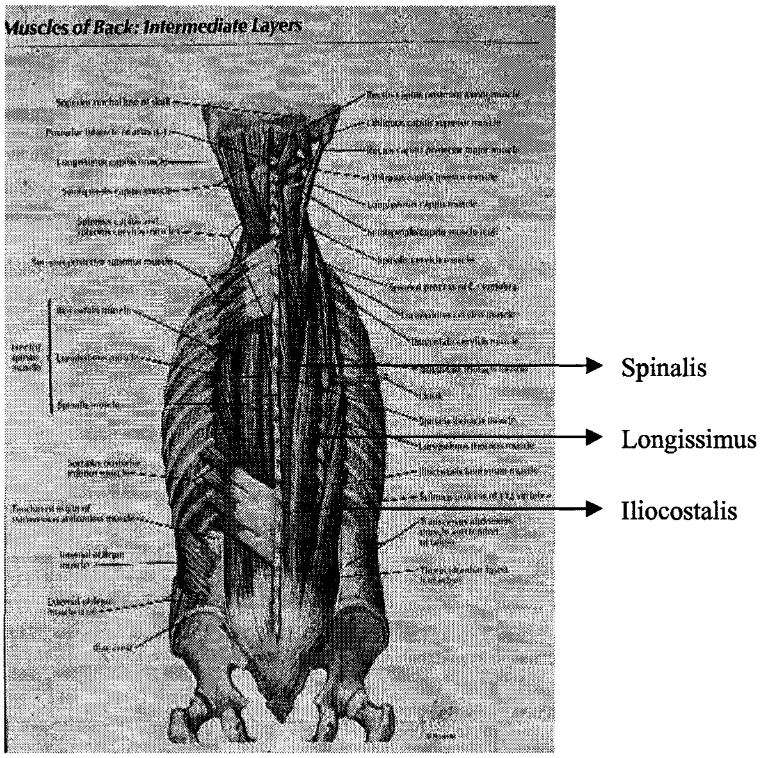

Figure 6: Posterior view of the three columns of the erector spinae.[61]

The erector spinae muscle group has been traditionally described as a common muscle mass that originates from the lumbar region and inserts into the lumbar and thoracic ribs and transverse processes.[62] However, more in depth research has identified that the erector spinae muscle group consists of 3 major divisions: spinalis, longissimus, and iliocostalis.[51] Each of these groups of muscle fibers have lumbar, thoracic, and cervical portions (example: iliocostalis lumborum, iliocostalis thoracis, iliocostalis cervicis). The common origin of the erector spinae musculature is via a broad tendon that attached inferiorly to the posterior aspect of the iliac crest, the posterior aspect of the sacrum, the sacroiliac ligaments, and the sacral and inferior lumbar spinous processes.[51] Specific anatomical descriptions will be outlined for the lumbar portions only. Iliocostalis is the most lateral of the erector spinae and its fibers run superiorly to the angles of lower ribs.[51] Longissimus is located between iliocostalis and spinalis muscle columns and its fibers run superiorly to ribs between tubercles.[51] Spinalis is the most medial erector spinae muscles and its fibers run superiorly to spinous processes in the upper thoracic region.[51]

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Contraction of these muscles works to extend the lumbar spine and also provide compressive forces to the lumbar vertebrae.[51]

Thoracolumbar fascia:

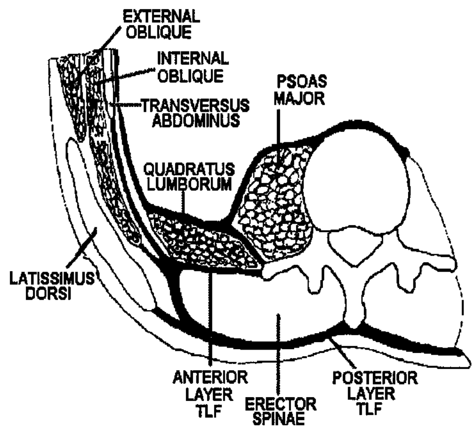

Figure 7: Transverse view of the left Thoracolumbar Fascia (darkened area).[63]

The thoracolumbar fascia (TLF) is described as consisting of three layers: the anterior layer (arising from the anterior surface of the lumbar transverse processes and covering quadratus lumborum), the middle layer (arising from the tips of the transverse processes and attaching to the TrA muscle), and a posterior layer (arising from the midline and covering the back muscles).[50] However, it has recently been discovered that the superior attachments of the posterior fascial layer are much more extensive than previously thought, exhibiting fascial continuity with the rhomboid and cervicis splenius muscles.[32] The role of the posterior layer of the thoracolumbar fascia has been viewed as generating increased tension to resist flexion moments in the spine and acting as a "posterior ligament" thus generating passive tension.[64] In this way, it has been described as a hydraulic amplifier, in that contraction of the back extensor muscles, when the fascial layer is taut, work to further increase the passive tension afforded by the thoracolumbar fascia.[65] Many studies have collectively questioned the idea that the thoracolumbar fascia could support substantial extensor moments.[64, 66] Therefore, the increased superior attachment of the posterior layer of the TFL may have significant implications to current computational models that might use the previously recognized attachments of the TFL that are less extensive. With the increased length of its superior attachments, the posterior

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layer of the TFL has the largest moment arm of all extensor tissues and the smallest potential for compressive forces on the spine.[64]

2.2 Theories of instability

Definition of instability

Instability has been defined by Panjabi[11] in terms of the "neutral zone", a region of laxity around the neutral resting position of a spinal segment. This can also be defined in biomechanical terms as the region where displacement can increase in a given direction without any appreciable increase in force. The neutral zone has been shown to increase with disc degeneration and injury[67, 68]and decrease with simulated muscle forces over a spinal segment.[52, 67-69] Panjabi theorized that *in vivo*, the neutral zone is influenced by 3 major subsystems: the passive (ligaments, vertebrae), active (muscles and tendons), and neural control subsystem (nerves and central nervous system).[11] Subsequentially, Panjabi defined clinical instability as a significant decrease in the capacity of the stabilizing systems of the spine to maintain intervertebral neutral zones within physiologic limits, so there is no major deformity, neurologic deficit, or incapacitating pain.[11] Using this definition, deficiencies in merely one or a combination of the subsystems could be responsible for the creation of instability in the lumbar spine.

It is important to note that instability can also occur during movements outside of the neutral zone, particularly when discussing dynamic stability in challenging postural environments. In this case, instability can be defined as the body's inability to maintain a posture or control a movement. For these situations, deficiencies in the same subsystems, as described below, can cause instability.

Role of the passive subsystem

It has been determined that the osteoligamentous lumbar spine can withstand merely 88 N of vertical force before lateral buckling of the spine occurs.[6] However, everyday activities present loads in the realm of 1,000 – 3,400 N [7, 8] with more extreme activities presenting in the range of 6,000-18,000 N.[9, 10] Therefore, it can be seen that the contribution of the passive subsystem to stability in the lumbar spine is relatively insignificant when the

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loads imposed upon the spine are analyzed. Rather, it appears to function as a framework to which muscles of the body can attach and stabilize through tension. As well, the passive subsystem can provide endpoints or restrictions to extreme movements.

Role of the active subsystem

Numerous hypotheses have been made regarding which muscles or muscle groups contribute most to spinal stability. Bergmark was the first to suggest that the muscles of the back can be divided into local and global muscles groups.[21] He hypothesized that the intersegmental or "local" muscles function primarily as stabilizers whereas the multisegmental or "global" muscles function primarily as motion producers.[21] However, research using computational models has suggested that no one single muscle has the main role of ensuring spinal stability; instead, all the muscles of the spine (both "local" and "global") work together collectively to maintain stability.[22, 23, 27, 65] A more in-depth look at both these hypotheses will be presented later in this chapter.

Role of the neural control subsystem:

Due to the relatively insignificant contribution of the passive subsystem to stability in the lumbar spine, the role of both the muscular and neural control subsystems in establishing stability becomes paramount. Neural control has been theorized to be responsible primarily when motor errors in a single vertebral segment occur. In an unprecedented study, injury to the lumbar spine was witnessed during fluoroscopic observation of a power lifter.[70] During the lift, the majority of the vertebral segments remained 2-3 degrees away from full flexion. However, one vertebral level demonstrated proportionately more flexion and a resultant buckling injury to the spine occurred. It was hypothesized that an inappropriate sequencing of muscular forces at this single level occurred due to a transient neural control error. These motor control errors have also been implicated in injuries occurring during low load situations[22] and during challenged breathing.[29] However, in order to fully understand the neural control and subsequent motor patterns required to maintain stability during functional activities, the contribution of muscular contraction to stability must be understood.

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2.3 Theories of muscular control

A. "Local" muscles – Mechanisms of stability

The first investigation into the muscular contributions to spinal stability was performed by Leonardo Da Vinci. Through his work with human cadavers, he hypothesized that the muscles of the cervical spine worked to stabilize the neck like the guy ropes on a ship's mast. He stated that "the more central muscles stabilize, the more lateral bend the neck".[71] Following this, an array of authors have concluded that muscles with segmental patterns of attachment are architecturally suited to producing the intersegmental spinal stiffness necessary to maintain stability in the lumbar spine.[21, 54, 68] It has been theorized that the intersegmental muscles play a key role in stability of the spine for the following reasons: First, the closer a muscle is to the center of rotation, the shorter the muscle length change will be for a given angular correction of the spine. Because there is a shorter muscle length change, this translates into a short reaction time and thus, a more efficient stabilization of the spine.[68] As well, it has been documented that the neural arch is not a rigid structure,[68] and that having the stabilizing musculature closer to the center of rotation removes the repetitive deformation of this arch every time the neuromuscular system makes a significant postural correction.[68] These postural corrections, if made by superficial, global muscles, such as the erector spinae, would result in the loss of muscular energy due to this deformation.[68]

Two specific muscles have been the primary focus of recent studies investigating the role of intersegmental muscles in ensuring spinal stability: the TrA and the multifidus muscle. These muscles are thought to have a primary function as stabilizers due to their unique, central, orientation that results in a very limited ability to produce trunk motion.[28] Furthermore, the anatomical make-up of these muscles suggests a stabilizing function. TrA, through insertions into the thoracolumbar fascia and subsequently, the lumbar vertebrae, is theorized to provide a corset-like stability to the spine when contracted. Multifidus, through its segmental fascicular attachments is speculated to act as a stabilizing factor during flexion activities, a common mechanism of injury for the lumbar spine. It is hypothesized that the co-contraction of TrA and multifidus has the potential to provide a dynamic corset for the lumbar spine, thus increasing its segmental stability

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during functional tasks[72] and providing justification for the "abdominal hollowing" exercise.

In support of the theory that spine stability is created through local musculature, several EMG studies have quantified the activation of TrA prior to upper limb [17] and lower limb movements[18] suggesting a proactive control of spinal stability. Specifically, the TrA has been shown to be active 30-100 ms prior to the activation of the prime mover in rapid arm or lower limb movement in a healthy population.[17, 18] Furthermore, these studies identified activation of TrA regardless of the direction of trunk movement,[52] the direction of the acceleration or deceleration of the trunk,[73] direction of limb movement,[17] or the direction of displacement of the center of mass.[14] These studies suggest that TrA contributes to spinal stability in a manner greater than simple control of spinal orientation. Further studies have shown that the pre-activation of TrA is both reduced in amplitude and delayed when experimentally induced muscle pain is introduced[16] and patients develop and adopt an alternate postural adjustment strategy when pain is present.[74] Based on the current theories in spinal mechanics, it is hypothesized that if changes in postural adjustments present in subjects with low back pain are sustained long term, they may pose a risk to spinal structures.[16, 23]

Studies on the lumbar multifidus have also implicated its role in lumbar stability, particularly through establishing its potential to provide control to vertebral segments within their neutral zones.[68, 69] Wilke et al. demonstrated that at L4/5, the multifidus muscle contributed 2/3 of the stiffness supplied by contraction of the muscles in close proximity to this level.[69] Recent research has shown that localized segmental dysfunction of the multifidus muscles occurs after a first episode of acute or subacute, unilateral low back pain.[13] Specifically, Hides et al. used ultrasonic imaging to demonstrate that multifidus atrophy occurred ipsilateral to the location of low back pain.[13] Furthermore, it has been shown that in a patient population with low back pain, multifidus muscle recovery did not occur in patients who did not receive a specific stabilization program even though their pain levels returned to normal.[12] However, in patients receiving treatment in the form of a specific stabilization exercise program, pain levels were

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reduced and full recovery of the multifidus muscles was present (as evidenced by increased multifidus cross-sectional area on ultrasound imaging).[12] Given these results, it was hypothesized that recurrence of low back pain may be influenced by this incomplete, non-automatic recovery of the multifidus muscle in patients not partaking in specific exercise training. Supporting this hypothesis, multifidus atrophy, as measured by MRI, has shown a correlation with leg pain in patients with low back dysfunction suggesting a relationship between low back dysfunction and the lumbar multifidus.[75]

Although the above studies implicate TrA and multifidus in the generation and maintenance of spinal stability, none of these studies have utilized quantitative testing to assess spinal stability directly. This omission limits the validity of the resulting conclusions. To the writer's knowledge, the only evidence for increasing spinal stiffness using a "local" muscle contraction, comes from a single animal study where electric stimulation was used to evoke a contraction of the TrA and diaphragm. During these contractions, spinal stability was quantified using a spinal stiffness measuring instrument that consisted of a posteriorly situated servocontrolled motor attached via a lever system to a pedicle screw within the target vertebrae.[15] These investigators measured increased intervertebral stiffness, giving support to the stabilizing role of TrA.[15] However, it is unclear whether results from external stimulation of a porcine model can be generalized to humans.

B. "Global"/Coordinated contraction of all muscles- Mechanisms of stability

Crisco and Panjabi[24] were some of the first investigators to study the effect that musculature architecture (e.g. intersegmental, multisegmental) has on the stability of the spine. Using a computational model, they determined that the multisegmental muscles (e.g. global muscles) were more efficient than the intersegmental muscles at stabilizing the spine to prevent buckling behavior in the frontal plane. Particularly, the pelvic-originating muscles were found to be 90% more efficient at laterally stabilizing the spine than the intersegmental muscles.[24] Furthermore, this efficiency was shown to increase as the muscle position became more laterally situated, [24] suggesting a more prominent role for the global muscles in achieving lateral spinal stability.[24]

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Building on this work, McGill suggested that no single muscle is more important in achieving spinal stability; rather that a coordinated activation of muscles (both local and global) is required to ensure stability during functional activities.[27, 65] Using the analogy of an upright and vertical fishing rod with guy-wires attached at different levels, he argued that each guy wire (muscle) needs to be of the same tension in order to sustain compressive forces successfully and ensure stable behavior in all directions.[65] It has been suggested that spinal stability depends on three conditions: 1) the symmetry of muscle stiffness and forces all around the spine, 2) the amount of co-contraction; 3) the geometry of the muscular guy wires (particularly the width of the base of support combined with their angle of pull).[65] As a result, it has been advocated that the "neutral" spine is required with equal isometric contraction of all trunk muscles to ensure maximal stability. The neutral spine has been described as the point at which the moment equilibrium is reached (back extensor muscles stop contracting) as a person extends from a slightly flexed trunk position.[65] Furthermore, in order to maintain a broad abdominal base, it was suggested that the patient should NOT "hollow in" the abdominal wall as this effectively decreases the size of the base of support, thus affecting the geometry of the muscular guy wires.[65] This theory was supported in research that established that modest levels of co-activation of the paraspinal and abdominal wall muscles while in an neutral position allowed sufficient stability of the lumbar spine to be achieved.[22, 23] In this work, sufficient stability has been defined as the amount of necessary muscle stiffness required for stability, together with a modest amount of extra stability to form a margin of safety [27].

Recent data suggest that all torso muscles exhibit perturbed onsets during sudden loading events in an athletic population.[76] Building on these findings, the theory of coordinated muscular actions suggests that by preferentially training the TrA, a faulty muscular pattern is being implemented.[65] Support is given for this theory through recent research by Kavcic et al.[26] who suggested that different muscles act as prime stabilizers depending on the particular functional movement. This research was performed using a computational analysis of normal, healthy patients performing various stabilization exercises that are commonly prescribed within a treatment environment.

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The majority of the studies performed supporting global contractions as stabilizing rely primarily on computational models and their analyses to determine the stabilization function of muscles. Limitations exist with these methods in that these models rely on several assumptions. Primarily, the main assumption is that muscular architecture and origins/insertions are comparable between individuals. However, for the muscle of the TrA alone, significant variation has been shown in the distal attachment of this muscle with complete or partial detachment of the muscle from the iliac crest and absence of fibers below the iliac crest.[31] This variability has also been documented in the lumbar erector spinae muscles and the thoracolumbar fascia.[10, 32, 77] The writer is not aware of any models completed to test these variations in anatomy. As a consequence, any calculations made where an assumed muscular origin and insertion are present remain global in nature and do not reflect individualized values. Additionally, studies have shown that that different individuals often utilize different muscle recruitment patterns as stabilization strategies when performing the same task.[23] Therefore, unless specific muscle activity and force levels are measured within individual subjects, a computational model would be further limited by the variability inherent within human subjects. Furthermore, computational models create numerical values of stability although, to this author's knowledge, these values have never been compared to a physical or surgical measure of instability for validation. Therefore, it is of utmost importance that stability is quantified within each individual instead of using a generic biomechanical model that does not account for individual differences. Using a stiffness assessment device that determines stiffness values globally, (regardless of muscular insertion locations - which eliminates anatomical assumptions made by current models), could perhaps reveal more information about the true value of stiffness.

C. Clinical prescription of stability exercises

It is apparent that if there is controversy regarding the "theory" behind instability and the muscular control of stability in the spine, there is also significant controversy regarding treatments designed to influence spinal stability. In the literature and in clinical practice, two main stabilization exercises are prescribed to patients who demonstrate symptoms of

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instability in the low back. The first exercise, "abdominal hollowing", is described to recruit TrA, and multifidus. This exercise is based on the theory that certain muscles act primarily as stabilizers of the spine.[34] Conversely, the second exercise, the "abdominal brace", is described to achieve co-contraction in the majority of the abdominal and back muscles with the primary goal being the maintenance of a neutral spine posture.[65] No focus on selective recruitment of any particular torso muscle is attempted in abdominal bracing. Clinically, both of the hollow and brace exercises can be integrated into treatment programs, although it is not clearly known which exercise more effectively increases spinal stability.

2.4 Comparison of stability: Abdominal hollowing versus bracing

Little scientific information exists which compares bracing and hollowing exercises directly. An early study, conducted by Richardson et al.[34] investigated the contribution of the "abdominal hollowing" exercise and the "abdominal brace" exercise to sacroiliac joint mechanics. Sacroiliac joint stiffness was assessed through a vibration analysis and was measured prior to and following implementation of the two stabilization exercises. Results of the study indicated that both the "draw-in" and the "brace" action increased stiffness of the sacroiliac joints, although the "draw-in" exercise had significantly more stabilizing effect on sacroiliac joint laxity than the brace. Specifically, abdominal hollowing decreased the laxity index from approximately 3 dB to below 1dB whereas the brace decreased the laxity value from approximately 2.5 dB to 1.25 dB, indicating a smaller increase in spinal stiffness generated by the brace as compared to abdominal hollowing.[34]

Conversely, Grenier and McGill, [25] used a computational analysis of the lumbar spine to determine a stability index that described the state of stability of the spine. When "abdominal hollowing" was implemented during different conditions (no load, load in both hands, right hand load, left hand load), a smaller stability index was found than that calculated to occur during the "abdominal brace". Therefore, the conclusion was made that the brace contraction was superior in providing stability to the spine. However, it is unknown whether the co-contraction of the lumbar multifidus muscle was used in this

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stability analysis, whose contraction could potentially increase the value of the stability index for the "abdominal hollowing" exercise. As noted above, this type of study carries the limitations of a computational model.

Most recently, a study examining the effect of abdominal stabilization maneuvers (compared the hollow and the brace) on the control of spine motion and stability against sudden trunk perturbations was completed.[78] Healthy subjects were posteriorly loaded in different experimental conditions: resting and no knowledge of perturbation timing; performing each of the stabilization maneuvers at 10%, 15%, and 20% of IO MVC with no knowledge of perturbation timing; and naturally coactivating the trunk muscles when perturbation timing was known. The results indicated that the hollowing maneuver was not effective for reducing the kinematic response (trunk motion) to sudden perturbation.[78] However, the bracing maneuver was significantly better at reducing trunk motion to sudden perturbation as compared to the hollow contraction; albeit at a cost of increasing spinal compression.[78]

2.5 Relationship between spinal stiffness and stability

Spinal stability itself has been documented as a very difficult concept to define, with previous studies noting the inadequacy of a wide-spread and commonly recognized definition.[79] Due to the nature of the spine, particularly the various movements it undergoes and loads that it disperses, it remains difficult to determine a specific test that would accurately measure the dynamic nature of stability that is required during functional tasks. As mentioned previously, muscular contribution to stability has been assumed to occur based on the timing and nature of muscular contractions,[17, 18] the use mathematical calculations and biomechanical models of the spine, [21, 22, 24, 28] and the use of assumed relationships of spinal stability to treatment outcomes (e.g. treatment success equals improved stability).[20] Also mentioned previously, limitations exist within all these methods, further supporting the documented difficulty in quantifying stability. This allows the findings of the present study to enhance the existing body of literature and improve the understanding of low back pain.

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Given the absence of an available method to quantify spinal stability, clinicians often use the posteroanterior (PA) pressure test to assess stiffness of the spine.[80] During this test, the clinician uses his/her hand to apply a downwards force to the spinous process of a vertebrae and attempts to subjectively judge how much resistance is given to his/her hand from the spine. Numerous authors have suggested that this test is very useful in identifying the spinal level to be treated as well as in determining a clinical diagnosis.[80, 81] Furthermore, the PA pressure test has been implicated as a measure to determine levels of hypermobility or instability within the spine.[80]

Recent research has supported the relationship of PA stiffness assessment to stability in the spine. Fritz et al.[82] found that one of the major predictors of radiological instability was a lack of hypomobility when testing PA stiffness in lumbar vertebral segments. Further, Hicks et al.[83] investigated the relationship between clinical variables and success or failure with a stabilization program. A positive prone instability test was found to be one of the predictors of success with a stabilization program.[83] This test is a modification of the PA pressure test whereby the examiner applies a posterior to anterior pressure to the lumbar spine (patient prone on the plinth with legs over the edge and feet resting on the floor) with any provocation of pain reported. Then the patient lifts the legs off the floor and the PA pressure is applied again. The test is positive if the pain is present in the resting position but subsides in the second position. Additionally, it was found that the absence of hypermobility during PA pressure tests was a significant predictor of failure with a stabilization program.[83] These findings suggest that a relationship does exist between the PA pressure test and stability in the spine.

It is recognized that the PA pressure test incorporates few aspects of timing, and is not performed in dynamic loading conditions, therefore, is not a complete measure of spinal stability. On the other hand, it can be argued that spinal stiffness is a representative measure of overall stability for two reasons: 1) Stiffness gives a value for the force needed to disrupt equilibrium, thus measuring a component of stability;[84, 85] 2) Disc degeneration (early stages), a predicted cause of increases to the neutral zone and thus, instability in the spine, has been shown to produce decreased levels of spinal stiffness as

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measured by a posteroanterior force.[86] In this way, justification for the use of spinal stiffness as a corollary of spinal stability can be made. The current methods by which spinal stiffness is quantified will be discussed in greater detail in the next three sections.

On a final note, recent literature acknowledges that stability cannot be defined within one context. [87] Primarily, it is suggested that the definition of stability is dependant both on the system and the task being performed.[87] It was suggested that two other features, robustness and performance, influence stability in dynamic conditions. In the case of the spine, robustness relates to how well the spine can cope with uncertainties and disturbances (using feedback control). [87] Therefore, with exercise, the spine as a system does not become more stable; rather, it becomes more robust. In this case, a stiffer spine will be displaced less than a compliant system and will respond faster, suggesting better performance. Therefore, increased trunk stiffness indicates that the spine is more robust than at lower stiffness levels in a static situation. However, depending on the activity, increased levels of trunk stiffness do not always produce a more robust spine. This brings in the third feature, performance, which reflects how closely and rapidly (accuracy and speed) the disturbed position of the system (the spine) tends to the undisturbed position.[87] Performance includes the reflexive and voluntary responses of the body reacting to different spinal positions. In this case, increased spinal stiffness by itself does not affect the activity of these reflexive responses, therefore does not always lead to better system performance. Further, increased stiffness is of the spine ***only*** not the entire system as is involved when performing an activity. Therefore, it can be seen that stiffness and stability do have a relationship, although that the aim should be to find the optimal level of trunk stiffness to ensure the task can be successfully completed.[87] In this thesis, the goal was to gain a better understanding of the effect of trunk muscle contraction on spinal mechanics within a static situation before attempting to address the complex aspects of neural control; therefore, using a static measure such as PA stiffness was justified.

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2.6 Manual methods of assessing spinal stiffness

Although relationships between the PA pressure test and patient response to a stabilization program have been shown to exist, limitations exist in the performance of this test. The primary drawback of this test is that it is subjective in nature. The accuracy of the test results relies on the ability of the human's tactile proficiencies to judge differences in stiffness. Accordingly, the PA pressure tests have been shown to be unreliable[88-90] and inaccurate in that they overestimate boney displacement and underestimate force applied.[91]

However, methods have been developed to improve the reliability of the PA pressure test. Specifically, an 11-point stiffness scale was developed that delineated stiffness into levels and a standard mechanical target (designated at normal stiffness on the rating scale) was implemented.[92] This improved the reliability of manual spinal stiffness measurements with the ICC value increasing to 0.55 (range 0.50 – 0.62).[92] The reliability was found to further increase (ICC = 0.77) with the addition of standard stiffness targets for each point on the 11-point stiffness rating scale and the use of a more rigorously controlled protocol.[92] However, the criterion-related validity of the PA pressures (as compared to the stiffness values found using the Stiffness Assessment Machine [93]) was low with an ICC of 0.56.[92] This was later improved when the examiners were given practice performing PA pressures at a force level and rate similar to the Stiffness Assessment Machine.[94]

Another method to manually assess stiffness involves the use of a modified handheld mechanical device to apply a high loading rate PA manipulative thrust to the spinous and transverse processes of lumbar vertebrae. The handheld device (Activator Adjusting Instrument) was equipped with a preload control frame and mechanical impedance head (load cell and accelerometer) allowing for measurement of both input force and acceleration response characteristics of the spine at the segment of contact.[95] However, reliability estimates were moderate with intra-subject variance reported to be 20-25%.[96]

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2.7 Alternative measurements of spinal stiffness

In order to combat the limitations and error imposed by a manual assessment of spinal stiffness, alternative methods of measuring spinal stiffness have been developed. Various research groups have used a variety of devices to quantify spinal stiffness. These are divided into three primary categories: 1) Use of automated indentation; 2) Use of vibration technology; and 3) Use of invasive technology.

A. Automated indentation or instrumented measurement of stiffness

Numerous researchers have employed this method to quantify spinal stiffness, although reliability and accuracy data is not available for all the various types of testing equipment. These devices work to collect both force and displacement data during a simulated central PA pressure test using a motor-driven indentation. Some commonly used instruments include: Spinal Physiotherapy Simulator (SPS), [97] Lee and Evans' stiffness assessment device,[98] Stiffness Assessment Machine (SAM), [44, 45, 93] Spinal Posteroanterior Mobilizer (SPAM), [99] and the Rigid Frame Indentor [100]. This section will further examine the design and methods of these devices.

The SPS measures stiffness by providing an oscillating force in a posterior to anterior direction, (controlled by a variable speed DC motor), at a frequency of 0.5 Hz.[97] A padded indentor is moved down via a parallelogram linkage system under the action of a dead weight. A load cell, situated in parallel with the indentor, is used to quantify force and two linear potentiometers are used to measure amount of skin surface movement. The first is mounted on the indentor and measures skin movement relative to the rigid indentor and second is mounted on the frame and measures motion of the indentor relative to the frame. Thus skin movement is the sum of the two potentiometer outputs.

Lee and Evans[98] developed a similar type device; however in this case, PA force is delivered to a selected vertebral segment while measuring the displacement of the skin over the adjacent vertebral segments. Therefore, the stiffness results are reported for a combination of levels (example: L3/4 and L4/5 values are reported when testing stiffness at L4).[98]

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In response to the large sizes and lack of portability of the previous two devices, the SAM[44, 45, 93] was designed. This instrument consists of a testing bed, a small metal pad (indentor) that applies the force to the subject, mechanical head (controls the movement of the indentor and measures the applied force and resultant displacement), and a control box (houses the analog-digital converter board and memory chips; is connected to laptop and used for data collection). The mechanical head is attached to a steel bar that sits over the test bed. The height of the steel bar above the bed surface can be adjusted to allow for assessment of patients of varying sizes. Further, the mechanical head also rotates about the bar allowing for the force to be applied in a cephalad or caudad direction (for testing different lumbar levels). A small reversible servomotor is stepped down using two pulleys to increase torque output and reduce speed (maximal force of 300 Newtons at 2 Hz). An inextensible cable travels around the lower pulley and attaches to the top of the indentor. Movement of the cable produces movement of the indentor pad. Displacement of the indentor is calculated by measuring the rotation of the pulley using an optical encoder and using this angular displacement to calculate linear displacement of the indentor. Force is indirectly measured using a conductive plastic linear potentiometer (using a spring, the potentiometer measures the equal and opposite force that is applied upwards by the patient's body in response to indentation). In this case, 1 mm of movement corresponds to 100 Newtons of force. The signal from the potentiometer inputs into the central processing unit and undergoes low pass filtering at 20 Hz and analog to digital conversion. Movement of the indentor is displacement controlled, such that the examiner specifies the desired amount of indentor movement using the computer menu. Preliminary loading cycles are undertaken to familiarize the patient and data collection is performed for 5 loading cycles to a max force of 105 N at a frequency of 0.5 Hz.[44, 45, 93]

In contrast, the SPAM[99] uses electrically driven load cell producing a loading rate of 0.583 mm/s and a max displacement of 15 mm. Load is applied to the spine via a rigid indentor consisting of dense foam padding. During stiffness testing, a pre-load of 30N is

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applied and then the load is applied for 10 seconds (maximum of 80 N). In this case, one measurement of stiffness is taken.

Similarly, a single measurement of stiffness is also taken with the Rigid Frame Indentation.[100] This system consists of a metal testing bed with a rigid metal frame that is situated over top of the test bed. A load cell in parallel with an ultrasonic indentor (A-mode ultrasound) quantifies force and a motor drives indentor movement. In this device, movement of the indentor can occur in both a lateral and anterior-posterior direction with downward displacement of the indentor measured through an optical encoder.

As it can be seen, there appears to be no set protocol in regards to type of loading (cycling at a set frequency versus taking one stiffness measurement). However, two outcome variables have been used most frequently when quantifying spinal stiffness using a PA pressure. The first is termed stiffness coefficient (K) and is calculated as the gradient of a regression line fitted to the force displacement curve between 30 and 90 N.[93] This range of applied force was chosen as it is the linear component of the force displacement curve and corresponds to the range of forces used by physiotherapists[101] during Grade II and II mobilizations in Maitland's grading system.[80] Displacement D30 is the second stiffness outcome measure and corresponds to the displacements in millimeters between 2 and 30 Newtons of force (the non-linear part of the force displacement curve) and occurs early in the loading range.[93]

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Reliability and accuracy values were found to be excellent for these devices (See Table 1); however, the clinical use of these instruments is limited by their large size, lack of transportability and/or lack of ease of use (difficulty testing patients that are unable to maneuver into the devices to attain the testing positions (prone lie).

	Reliability	**Accuracy**
SPS[97].	ICC = 0.88 at L3	Underestimated true stiffness of elastic beams by less than 1%
Lee and Evans[98].	ICC = 0.99 at L3/4 ICC = 0.95 at L4/5	Unknown
SPAM[99]	ICC = 0.979 at L5	Standard Measurement Error = 0.515 N/mm
SAM[44, 45, 93]	Test-re-test: ICC = 0.96 for	Underestimated true stiffness of elastic beams at a maximum of 2.5%
Rigid Frame Indentation[100]	ICC = 0.99 – 1.00 over varying experimental conditions	Error values of 0.81% - 13.62% over varying experimental conditions

Table 1: Reliability and Accuracy Values for Automated Indentation Devices

B. Vibration technology

Stiffness has also been quantified in other forms that do not involve using a PA pressure test. One such way is using Doppler imaging of vibrations (DIV). Often vibrations are used to assess stiffness, as vibration transmits better between joints with increased stiffness. Assessment of the transmission of vibration is often measured using color Doppler ultrasound.

A device called the Vibrator Derritron VPE (Derritron Electronics Ltd., Hastings, East Sussex, England) was used in a study by Richardson et al. to test the stiffness of the sacroiliac joint.[59] In this system, vibrations were applied unilaterally to the anterior superior iliac spine of patients and then resultant transmission of vibration on both the sacrum and ilium was assessed using color Doppler imaging transducer (covering both sides of the joint). Colored pixels resulting from the vibration of the sacrum and ilium appeared on the monitor at high threshold values (dimension decibels). Then, threshold values were calculated for both the sacrum and the ilium (where the color image

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disappeared and changed to a gray scale). Because the threshold value is directly related to the vibration velocity of the bone, a large difference between the threshold values of the ilium and sacrum represents a large difference in vibration velocities and thus a low value of stiffness. Intra-rater reliability has been shown to be good to excellent with ICC values ranging from 0.75-0.89 for experienced testers and accuracy is limited to changes in threshold values larger than 1.45 - 2.38.[102]

C. Invasive measurement

Quantification of spinal stiffness using invasive methods has been used primarily in the porcine model. Hodges and Kaigle [15] tested porcine spinal stiffness via a torque applied by a servocontrolled motor to the L4 vertebrae via a level system attached to pedicle screws and a cross-bar. The motor was attached to a lever, which through a multi-axial joint, was attached to the crossbar. In this way, the motor displaced the distal end of the lever both caudally and rostrally 2 cm and thus this torque was applied to the vertebrae. The torque was applied using a frequency of 0.5 Hz for 4 cycles with the force required to displace the vertebrae measured using a force transducer in series with the lever system. An electro-magnetic motion analysis system quantified displacement of the vertebrae by tracking motion sensors via intraosseous pins in the L3 and L4 spinous processes. As per above, the linear region of the force displacement curve was used to estimate spinal stiffness. Reliability of this device has not been tested. However, it is apparent that this type of approach, due to the significantly invasive nature, is only appropriate for use in animal models.

2.8 Assisted indentation:

Assisted indentation (AI) is a non-invasive procedure that addresses many of the issues raised by manual assessment of spinal stiffness through objective quantification of the force displacement properties of spinal tissue. (See Figure 8) Further, it helps address the transportability issue inherent in many of the automated stiffness measurement devices. In this device, movement of a cylindrical rod simulates a posterior-anterior digital compression force used in manual palpation. The indentor is manually displaced by the researcher. During the PA-indentation, the parameters of force and boney displacement

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are quantified by AI using a load cell in series with the indentor and a displacement transducer parallel to the indentor. This allows for an objective measure of spinal stiffness within each individual subject.[35] In this way, limitations in accuracy of spinal stiffness estimation inherent in the manual subjective assessment of force provided and boney displacement are minimized.

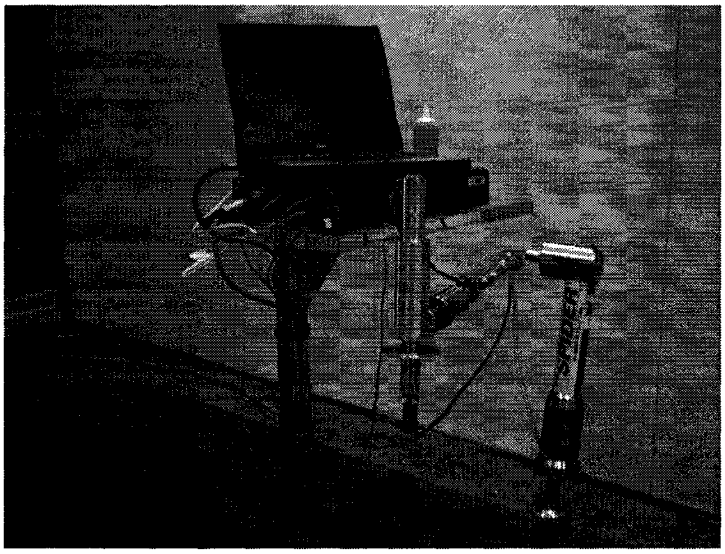

Figure 8: Assisted Indentor

This form of AI has demonstrated good reliability and accuracy in the evaluation of stiffness and boney displacement measures within a force-displacement model of the lumbar spine.[103] Overall accuracy of the AI technique has been demonstrated to be 0.04 mm. As well, the accuracy of assisted and rigid frame indentation (as described in the previous section) was not significantly different ($p = 0.083$), demonstrating that these techniques are comparable.[103] Furthermore, due to its small size and relative ease of transportability, the assisted technique of indentation is well-suited for use in a human population. Although the accuracy has been established earlier by Kawchuk et al., [103] little testing has been completed on the reliability of this device in-vivo. Therefore, the reliability of this researcher was established in this area prior to use. This will be discussed in further detail in Chapter 3.

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2.9 Factors affecting measured stiffness values

Numerous factors exist that can influence measured stiffness values within a human population. These factors can be sectioned into two major categories, with the first being subject-related factors. These factors include patient movement during testing, variation in intra-abdominal pressure (stiffness measures taken during different cycles of breathing), and muscle contraction.[35] Also included within this category is subject anxiety and relaxation. When measuring stiffness within a person, the first trials may exhibit higher stiffness values, and this phenomenon may be directly related to the subject's ability to relax (level of anxiety). Then, in subsequent trials, stiffness values can decrease, due to relaxation of the patient. Overall, these subject-related factors have the potential to alter the measured stiffness value of a tissue and unless controlled, can decrease the reliability of stiffness measures.

The second category of factors that affect measured stiffness is related directly to the procedures used during stiffness testing. It is well documented that the rate of indentation influences stiffness values, with higher rates of indentation producing increased levels of stiffness.[37] Therefore, fluctuations or changes in indentation rate directly alter measured stiffness values. Further, stiffness testing at higher rates can cause reflex reactions within muscles, causing a significant increase in the stiffness values.[104] Lastly, the time between stiffness measurements affects the measured stiffness values primarily due to the viscoelastic nature of tissues in the human body. With application of a load to viscoelastic tissue, increased fluid flow is promoted.[37] If there is insufficient time between stiffness measurements, there may not be enough time for the fluid to return to the tissues, resulting in increased levels of measured stiffness at subsequent indentations. This has been demonstrated in previous studies.[59] However, stiffness measures that utilize consecutive stiffness measurements (example, 5 indentations at a frequency of 2 Hz) have also been implicated in causing a treatment effect (reducing stiffness), particularly in situations where both joint and soft tissue is being loaded.[80]

Therefore, it can be seen that many factors can influence measured stiffness values. However, the effect of these factors can be minimized through robust standardization. In

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the present study, these factors, both subject-related and procedure-related were standardized and performed in a similar manner in each participating subject. Further detail is given in Chapter Three.

2.10 Surface electromyography

A. Factors affecting the validity of surface electromyography

Surface electromyography (sEMG) is commonly used to quantify the level of muscle activity, particularly due to its convenience and non-invasive nature. Electromyography measures muscle function by detecting and measuring the bioelectric signals that occur in innervated muscle.[105] In EMG, the functional unit of contraction is considered the motor unit, consisting of an alpha motor neuron and the muscle fibers it innervates.[105] These muscle fibers will only contract when the action potential of a motor nerve reaches its threshold. This depolarization generates an electromagnetic field which is detected by EMG and expressed as a voltage.[105] Therefore, EMG recording is a measure of the summation of the motor unit potentials within the effective "pick-up" area of the surface electrode.[105]

Because electrodes are placed on the surface of the skin and not directly into the muscle, as in the case of fine-wire EMG, many factors can work to confound surface measurements of muscle activity. Even using the proper skin preparation (shaving area to remove any hair and dead skin as well as cleaning with an alcohol swab), [106] problems with the acquisition of sEMG can still occur. One primary limitation of sEMG is the occurrence of crosstalk, or the interference of the EMG signals from adjacent or deeper muscles that are within the pick up area of the electrode. However, there are many ways to combat this problem, such as using smaller electrodes,[107] ensuring accurate placement of the electrodes,[107, 108] using differential electrode configurations,[109, 110] and monitoring subject characteristics (for example, body composition).[107, 111]

It is recommended to use as small as electrode as possible as the probability of crosstalk occurring increases with greater width and length of the detection surfaces as well as with increased inter-electrode distance due to increased pick-up area of the electrodes.[107] As

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mentioned above, accurate placement of the electrodes is important to minimize the effects of unwanted crosstalk. It is reported that electrodes should be placed between a motor unit and the tendon insertion, or between two motor points, along the longitudinal midline of the muscle[107] parallel to the muscle fibers.[108] Further, it has been shown that placing electrodes over the motor point results in a more unstable EMG signal due to the increased neural density at this point.[108]

Furthermore, using a bipolar differential configuration (where two electrodes detect two separate potentials within the muscle of interest relative to the reference electrode) works to decrease the effect of crosstalk. In this case, the signals are fed into a differential amplifier and then the difference between the two signals is amplified. Thus, this configuration serves as a bandpass filter whose bandwidth is a function of the spacing between the electrode surfaces and the common signal is removed.[109] Double differential configuration (where three electrodes detect separate potentials within a muscle) has been shown to be the most effective at minimizing the effects of crosstalk.[110] Also, in order to reduce overall noise it is recommended to secure the wires that connect the electrodes to the main unit so that there is no bending or pulling on the wires that can create a noise artifact.[112]

Although many of the above factors, are dependent on the equipment or set-up of the EMG, patient related factors also exist that can influence the presence of crosstalk. It has been reported that crosstalk from adjacent muscles increased with the size of the subcutaneous fat layer.[107, 111] Therefore, having subjects with less adipose tissue would be beneficial to achieve the most valid signals from muscles. Lastly, when using sEMG it has been suggested that manual resistive techniques should be used to test isolated muscle contraction to determine whether the electrodes have been placed properly on the muscle and connected to the equipment so that a reliable sEMG signal can be recorded.[108]

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B. Reliability of sEMG

In terms of reliability, sEMG has been shown to be site dependent, particularly when testing the abdominal musculature. A common electrode site measuring internal obliques and TrA demonstrated good reliability between days with an ICC value of 0.90 for unilateral shoulder flexion and extension.[113] However, the external oblique and rectus abdominis sites both demonstrated much lower reliability, possibly due to the larger levels of adipose tissue around these areas, which may affect signal integrity.[107, 111] A study assessing EMG activity of the erector spinae muscles in quiet stance, demonstrated very good reliability of all EMG signals (ICC > 0.75).[114]

The above reliability values discuss the repeatability of sEMG within subjects. In order to compare EMG values between subjects, normalization techniques must occur whereby the amount of muscle contraction obtained is expressed as a percentage of the subject's maximal voluntary contraction (MVC). In a study by Dankaerts et al., MVC of the abdominal and back musculature showed excellent within-day reliability for healthy controls with ICC values ranging from 0.87-0.98 and standard error values as a percentage of the grand mean of 4%.[115] This suggests that using an MVC contraction to normalize EMG data is a reliable process when using standardized testing positions for maximal muscle activity.

2.11 B-mode ultrasound

A. Use of B-mode ultrasound to quantify TrA contraction

B-mode ultrasound is being used with increasing frequency to assess and diagnose different medical conditions and pathologies. This is due to its low-cost and absence of radiation, as well as its ease of use and accessibility as compared to magnetic resonance imaging (MRI). Particularly in the situation of attempting to observe and quantify deeper muscle function, B-mode ultrasound has been used as a replacement for more invasive technologies such as fine-wire EMG.

It is well-known that changes in both muscle geometry and shape, such as muscle pennation angle and fiber length) occur with muscle contraction.[14, 116, 117] Furthermore, it

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has been shown that a curvilinear relationship exists between contraction level and changes in these parameters for both trunk and limb muscles, [116-119] but for contractions below approximately 30% of a maximal voluntary contraction (MVC), the relationship becomes more linear.[118] More specifically, a linear relationship between changes in muscle thickness of TrA and contraction level up to 30% MVC have been shown to be present.[118] Thus, measurement of changes in muscle thickness of this muscle provides a viable option to measure the function of the deep trunk muscles such as TrA.

Numerous studies have investigated the use of real-time B-mode ultrasound imaging in measuring TrA function. These assessments of muscle function usually involve measuring the thickness of TrA (measured between the superficial and deep borders of the muscle as visualized by the hyperechoic fascial lines) at two or more locations and comparing the thickness of the muscle at rest to the thickness during contraction. In this way the change in thickness values produced with a particular exercise has been used as a measure of muscle function.[118, 120-124] Additionally, certain groups have used normalization techniques to express the change in TrA thickness as a percentage of the maximal voluntary contraction of TrA (using a Valsalva maneuver).[118, 122] However, image visualization of TrA during maximal contraction is often impaired at full MVC, thus some groups advocated using 50% MVC.[118]

B. Reliability and validity of B-mode ultrasound in TrA contraction

In order to quantify muscle function using ultrasound, many factors need to exhibit reliability. Studies of the measurement procedures for the use of B-mode ultrasound to quantify TrA contraction level have shown excellent results. Intra-image, intra-rater reliability was found to have an intra-class correlation value (ICC) of 0.98 and an inter-image, intra-rater reliability ICC value of 0.93.[123] In this situation, the rater's ability to identify and measure thickness on the same image was tested (intra-image, intra-rater reliability). However, more importantly, measurements from two separate ultrasound images (both while the subject was resting) were found to be reliable (inter-image, intra-rater reliability). This encompasses both measurement reliability (TrA thickness) as well as reliability in the procedures used to obtain standardized image locations. In another

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study, testing in a variety of patient positions (sitting, standing) produced excellent ICC values for the measurement of TrA thickness between trials on a single day, ranging from 0.90 – 0.96 (SEM 0.29 – 0.57 mm).[125] When comparing between days, the ICC values remained high, with standing positions having an ICC of above 0.96 and sitting of 0.88 (SEM 0.18-0.33mm).[125]This suggests that excellent reliability (regardless of patient position) is possible if proper standardization techniques for acquisition and measurement of ultrasound images are utilized.

Recent studies suggest that the measurement of TrA contraction using B-mode ultrasound is also a valid technique. Specifically, the concurrent validity of this technique in the measurement of muscle function has been established. A study by McMeeken et al.[122] demonstrated that changes in TrA muscle thickness measured by B-mode ultrasound exhibited strong correlations ($R^2 = 0.87$, $p < 0.001$) with fine-wire EMG output from the TrA, suggesting that this method provides a valid measurement of muscular activity. Further, thickness measures of the TrA using ultrasound images were also found to correlate well (ICC = 0.78 to 0.95) with thickness measurements using MRI.[126]

C. Use of cross-sectional area to quantify TrA contraction

Due to the requirements of Experiment Three in this thesis (patient situated in prone lying to allow measurement of spinal stiffness), the contraction of TrA using B-mode ultrasound had to be visualized posteriorly rather than anteriorly. This posterolateral approach is not common, but was used here because of the prone nature of the patient. Because of the use of this imaging angle, the resulting ultrasound images were different than that which would have been obtained in the traditional anterolateral approach. Specifically, posterior visualization of TrA altered the orientation of TrA on the ultrasound image such that it was not always horizontal (not parallel to the borders of the ultrasound image), but instead was angulated. Second, TrA often exhibited a curved nature in ultrasound images. (See Figure 9)

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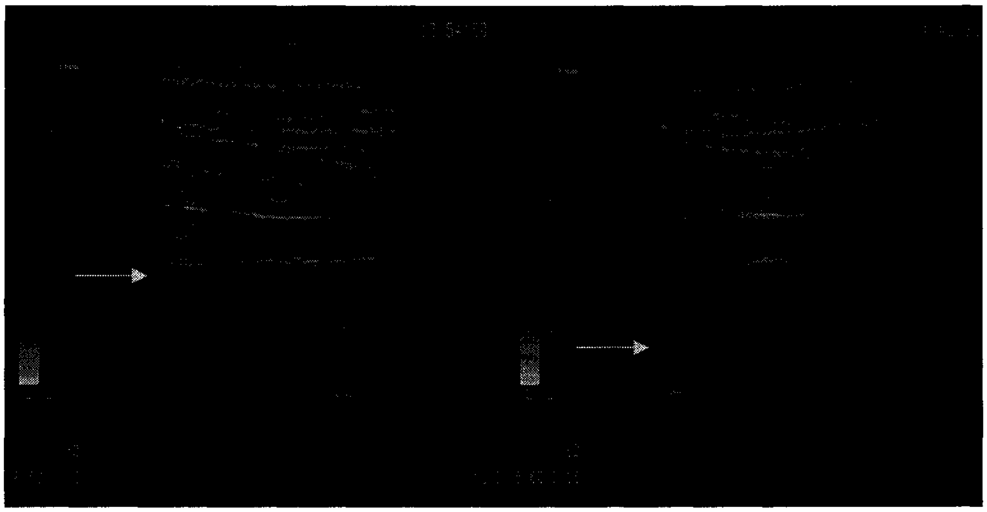

Figure 9: Ultrasound pictures taken during performance of the brace; left picture – rest, right picture – brace. TrA is horizontal on the screen, making thickness easy to measure (left picture, yellow arrow); TrA is diagonal on the screen with some curving (right picture, yellow arrow).

As mentioned previously, when imaging the anterior aspect of TrA, thickness measurements are used to quantify muscle function.[59, 118, 121-124, 126, 127] However, the demonstrated reliability of thickness measures of TrA is representative only of anterior imaging of this muscle, where the TrA is positioned horizontally on the ultrasound image. As it can be seen in Figure 9, this horizontal position was rarely maintained during contraction of TrA. Further, with curvature of the TrA muscle, thickness becomes more difficult to measure as it must be performed perpendicular to the muscle. If the measure of thickness is not perpendicular, a falsely increased value of thickness could be obtained. Therefore, if using only one measure of thickness, the reliability of this technique may decrease.

Based on the differences between ultrasound images that occurred due to the use of two imaging approaches, traditional measurements may be inappropriate to measure the images obtained from posterior visualization. Specifically, due to the unique curving of the TrA muscle in posterior visualization, CSA may be more reliable than thickness measures as more of the muscle bulk is taken into consideration, rather than one discrete portion. Therefore, the CSA method of quantifying TrA contraction was developed (See Method Section for Experiment Two). It should be noted that CSA measurement on B-

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mode ultrasound images has previously been used to quantify contraction of the multifidus muscle.[128] Due to the anatomical configuration of the multifidus muscle, the CSA of this muscle, rather than thickness, is used to quantify changes in muscle bulk that occur with contraction. Intra-rater reliability of the measurement of multifidus CSA has been determined to be excellent with ICC values ranging from 0.98 and 1.00. The 95% limits of agreement for between-scans reliability was approximately -0.25 to 0.5 cm^2.[128] Additionally, the validity of this method has been established in previous studies where multifidus CSA measurements made using ultrasound imaging were not different than CSA measurements made using magnetic resonance imaging (MRI), a gold standard.[129]

Although previous literature does support the use of this CSA measurement, because a different location was used (multifidus not TrA), the reliability and validity testing of this method of CSA measurement of TrA contraction was performed in Experiment Two. However, in order to compare TrA contraction findings with previous studies using TrA thickness, an average thickness value was also calculated based on the CSA measured and the length of TrA muscle over which CSA was taken.

Summary

Based on the literature, it is apparent that there is a lack of knowledge regarding the relationship between muscular contraction and stability in the lumbar spine. There is a need to understand the effects that changes in muscular contraction (exercise performance) have on the stiffness of the spine. Further, it has been shown that reliable and accurate tools are available that can quantify alterations in stiffness as well as measure muscle function. For those tools whose reliability and accuracy are unknown, experiments will be performed to determine these properties.

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CHAPTER THREE
METHODS

Overview

This project consisted of three experiments that assessed 1) the intra-rater reliability of AI, 2) the intra-rater reliability and concurrent validity of cross-sectional area measurement of B-mode ultrasound images of TrA, and 3) the effect of two abdominal muscle contraction patterns on spinal stiffness. The methods for each of these experiments are described below.

3.A. Experiment One: *Intra-rater reliability of assisted indentation*

This experiment was designed to measure the intra-rater reliability of a single rater in measuring spinal stiffness within a human population. It was hypothesized in this experiment that the intra-rater reliability would be excellent (ICC ≥ 0.75) where an ICC value ≥ 0.75 is considered to be "excellent".[130]

3.A.1 Subjects

Subjects were asymptomatic male and female volunteers who were recruited primarily from 1) staff at the University of Alberta, 2) students attending the University of Alberta, and 3) referrals from previous studies.

3.A.2 Sample size

It was calculated that a total of 23 subjects were required to yield a power of 80% toward detecting a difference in stiffness measures between indentations when a difference truly exists. This was based on a significance level of 0.05. (See Appendix B for sample size calculations).

3.A.3 Inclusion criteria

The subjects who participated in this study were asymptomatic subjects between the ages of 18-30 without a history of low back pain within the last year as well as an absence of

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current low back pain. Thirty years of age was chosen as the upper limit as it has been demonstrated that the incidence of disc degeneration may increase beyond this age.[131] For a more detailed description of the inclusion criteria, please refer to the methods section pertaining to Experiment Three.

3.A.4 Exclusion criteria

Subjects were excluded from this study if they reported back pain and/or medical conditions that could affect the safety of measurement of spinal stiffness using AI. Please refer to Table 2 for a full list of exclusion criteria.

Injury related	Disease processes.[42]	Subject factors
Current low back pain	Osteoporosis	Pregnancy (unsure or confirmed)
Low back pain within the last year	Osteoarthritis	Medications affecting muscle function (e.g. steroids)
Previous back surgery	Rheumatoid arthritis	Medications affecting pain recognition (e.g. pain medications)
Lower extremity injury within the last year	Ankylosing Spondylitis	Unable to tolerate indentation
	Known malignancy	
	Known spondylolisthesis	
	Multiple sclerosis	
	Severe scoliosis	

Table 2: Exclusion criteria.

3.A.5 Data collection

Data collection and analysis were conducted in the Common Spinal Disorders Lab in Corbett Hall (CH 3-44, University of Alberta).

3.A.6 Study protocol

With the subject lying in a prone position on a padded plinth, the subject's spine was palpated by the researcher and the L4 spinous process was identified. Although identification of spinous processes in the lumbar spine has demonstrated moderate

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accuracy with use of preferred palpation procedures (47% were on the level intended);[132] a standardized procedure was utilized in this study to identify the L4 spinous process in all subjects. Specifically, the horizontal line between the iliac crests was used to identify the L4/5 interspace and the vertebrae above was determined to be L4 (if this line between the iliac crests gave a spinous process, this was identified as L4). [133] The skin over this landmark was marked using a pen to provide a visual guide for placing the indenter. The indenter was then placed over the ink marking to provide a series of five consecutive indentations to familiarize subjects with the indentation process. Once the familiarization indentations were completed, 10 consecutive spinal stiffness measurements (indentations) were collected, each separated by a time period of two minutes. During times between indentations, subjects were instructed to remain in a resting prone position and were instructed to remain stationary and relaxed. Indentations were performed by one researcher (TL) who had logged approximately 100 hours of using the indentation device prior to data collected for this experiment.

3.A.7 Measurement of spinal stiffness: Assisted indentation

Instrument:

The equipment used for the indentation testing was designed and fabricated at the University of Alberta. A more detailed description of this instrument has been published previously.[103] In brief, the indentor consisted of a main outer frame and an inner cylindrical probe used to apply force to the anatomical target of interest. A compressive-tension load cell (Entran, Fairfield, NJ) was connected in series with the cylindrical probe with a screw-in mechanism located at the proximal end of the probe. By employing a circular aluminum platform connected to the probe, the researcher's hand could advance the probe downward to exert an indentation force. Because the displacement of the indenter is initiated by a manual process, but restricted by mechanical boundaries, this form of indentation is called "Assisted Indentation". By using a ceramic air-bearing to hold the indenting probe, frictionless movement of the inner probe with respect to the outer frame was achieved thereby reducing artifacts due to movement of the frame during indentation loading. A displacement sensor (LVDT) attached to the aluminum platform and the outer frame was used to measure total indentation displacement. This entire

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device was attached to a pneumatic surgical arm that allowed the indenter to be moved then locked into position. Electronic signals from the load cell and the LVDT were conditioned appropriately and collected by customized LABview software at a collection rate of 200 Hz.

Calibration:

Calibration of the assisted indention device was achieved using calibration masses of known magnitude applied to the load cell and spacers of known dimensions applied to the LVDT. After each application of increasing calibration mass or dimension, force and displacement signals were collected then plotted against the known mass or dimension. These data were then modeled with a linear curve. In each case, the r^2 value of the line of best fit was greater than 0.90. The resulting equation of the line of best fit was then used to determine the units of measure for the output voltage of each transducer. Calibration was completed prior to the testing of each subject.

Spinal stiffness measurement:

In each prone subject, the AI device was placed perpendicular to the L4 spinous process with a contact load of less than 0.1 Volts. The subject was then instructed to breathe out comfortably then to hold his/her breath for the duration of the indentation (approximately 5 seconds).[35] During indentation, the indentation probe was advanced manually (approximately 2 mm/sec) into the spine until a force threshold of 100N was read from a visual indicator. This level of force application has been shown to be safe for included research subjects as forces up to 200N have been used within a human population without any adverse effects reported.[43] When the 100N threshold was reached, the indentor position was maintained for approximately 1 second after which the indenter was removed from contacting the subject. During indentation, the equipment operator used a second visual display of indentation rate to decrease variability in the desired indentation rate of 2 mm/sec.

During the indentation process, all subjects held an analog trigger to indicate if their level of discomfort during indentation increased from baseline. If the subject wanted

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indentation to cease for any reason, at any time, they were instructed to squeeze the trigger fully which produced an audible alarm alerting the researcher to remove the indentor. If this situation occurred, the researcher re-positioned the indentor and indentation was attempted again as per guidelines outlined in the approved ethics application. Re-positioning of the indentor was allowed a maximum of two times after which further indications of painful indentation excluded the subject from further participation.

Indentation data (force and displacement) were used to calculate the spinal stiffness at the indentation site. Stiffness was quantified in two ways: 1) Global Stiffness (GS); and 2) Mean Maximal Stiffness (MMS). Global stiffness was calculated as the slope of the force displacement curve between 30 N and maximal force and it represented the stiffness of the vertebrae during the indentation itself. (See Figure 10) Mean maximal stiffness, the second variable representing stiffness, is the average stiffness value (N/mm) taken over the time period where the maximal indentation force has been delivered then held for a period of approximately 1 second. (See Figure 11) The MMS variable is therefore a ratio between the applied force and the resultant displacement of the underlying tissues.

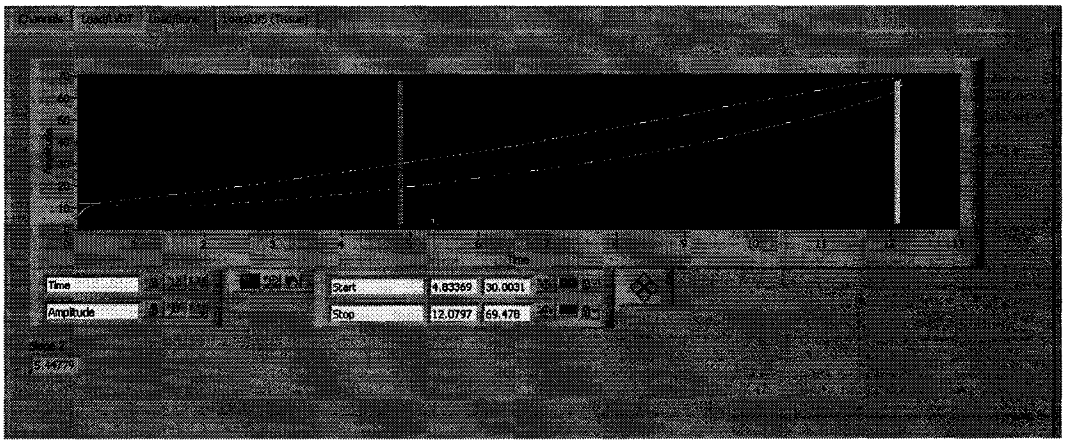

Figure 10: Output data for GS. Force values (N) are on the Y-axis with displacement data (mm) on the X-axis. Slope is taken from the green line to the yellow vertical line.

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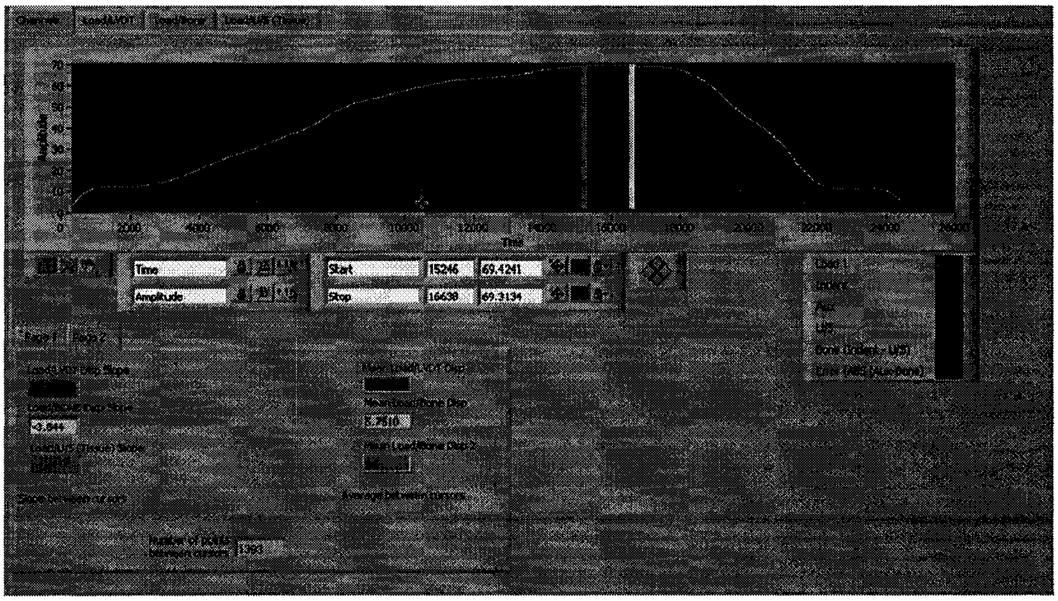

Figure 11: Output data for MMS. Amplitude values (N or mm) are on the Y-axis with time (sec) on the X-axis. Stiffness values are calculated for all points (approx. 1500) between the green and yellow vertical line. These values are then averaged to calculate mean maximal stiffness.

3.A.8 Statistical analysis

For data analysis purposes, all five of the familiarization trials were discarded. In addition, the first trial (stiffness measurement during rest) of the ten experimental indentations was discarded as this first trial has been shown to highly variable [45, 93] while stiffness measurements from subsequent trials (after the first trial) have demonstrated stability.[45, 93]

To assess intra-rater reliability of the researcher/instrument in measuring spinal stiffness, the intra-class correlation coefficient (3,1) was calculated. Shrout and Fleiss described three models of the intra-class correlation (corresponds to the first number in the parentheses).[134] In Model 3, each subject is assessed by the same rater(s), but the rater(s) represent the only rater(s) of interest.[134] Further, the second number in the parentheses signifies the form of measurement, whereby either a single measurement is used (represented by the integer 1) or the mean of several measurements are used (represented by the letter k) as the unit of analysis.[134] In the present study, intra-rater reliability was not assessed using a test/re-test model. Instead, this study assessed reliability via a single rater and her repeated measurement of stiffness in a single subject.

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To further describe reliability, inter-trial inconsistency (ITI) values for stiffness variables were calculated by taking the difference between two consecutive indentations expressed as a percentage of the average of the same two indentations.

Finally, to further explore reliability and investigate the possibility that a gradual change in stiffness values may occur with successive indentations, a condition that may not be reflected in ICC values, two test were used. The first, Pearson's r, was used to determine the correlation between each indentation (2-10) and second, paired t-tests were used to determine if a significant difference in stiffness outcome variables was present between the 2^{nd} and 10^{th} stiffness measurements.

3.B Experiment Two: ***1) Intra-rater reliability of cross-sectional area measurement of TrA on B-mode ultrasound images; 2) Concurrent validity of CSA measurement of TrA.***

This experiment had two goals. First, this experiment was designed to determine the intra-rater reliability of cross-sectional area measurements of TrA using B-mode ultrasound. It was hypothesized that reliability of one rater at measuring TrA CSA would be excellent (ICC ≥ 0.75).[130] The experiment's second goal was to ascertain the validity of CSA measurements of TrA by calculating the TrA activation ratio (TrA size while contracted/TrA size at rest) and then comparing these results to published activation ratio values.[123] The activation ratio reflects the size increase of TrA during contraction that is expected to occur with trained subjects. A comparison of the percent error of both thickness and CSA measurements in situations where potential measuring error could occur was also completed.

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3.B.1 Subjects and sampling

For the reliability component of this experiment, results from ten subjects of Experiment Three were chosen randomly. Specifically, pairs of ultrasound images were available from each of the following conditions: rest, hollow, or brace. From these, a random image pair was selected for 10 subjects. As a result, a total of 20 images (10 subjects * 2 images per condition * 1 condition per subject) were used for reliability analysis.

For the validity component of the experiment, ultrasound image pairs for all subjects in Experiment Three for the rest and hollow condition were utilized. This resulted in a total of 224 images (28 subjects * 2 images for rest * 2 images for hollow * 2 repetitions of each of the contractions) for the validity analysis.

Lastly, one image of a TrA contraction that exhibited significant angulation and curvature of the muscle (factors which were thought to affect the reliability of traditional thickness measures) was used to compare the percent error of thickness and CSA measurements.

3.B.2 Inclusion criteria/Exclusion criteria

This study involved the re-analysis of subject data from Experiment Three. Please refer to the methods section of Experiment Three for more detailed description of inclusion and exclusion criteria.

3.B.3 Data analysis

Data analysis occurred at the Common Spinal Disorders Lab (3-44 Corbett Hall, University of Alberta).

3.B.4 Study procedure

For the reliability study, the cross-sectional area of TrA was measured on ten ultrasound image pairs (10 subjects * 2 images per subject * 1 condition per subject) by one rater on the same day. Measurements were separated by a time of 4 hours and files were re-named by an independent researcher to prevent recall.

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For the validity study, the mean of the ultrasound image data for the rest and hollow contractions (as analyzed in Experiment Three) for each individual subject was used to calculate a TrA activation ratio.[123]

$$\text{Activation ratio of TrA}^{123} = \frac{\text{Size of TrA contracted}}{\text{Size of TrA at rest}}$$

For percent error calculations, the effect of error in drawing a line perpendicular to the muscle bulk of TrA (as done with thickness measurements) was examined with three cases of error investigated. Specifically, one line was drawn at an angle of 4 degrees "off" of the original perpendicular line on TrA, another at 13.5 degrees "off", and the last at 21.1 degrees "off". Four measurements of thickness and CSA were made. One measurement was termed "normal" in that the thickness and CSA were measured over the proper line bisecting TrA (perpendicular to the muscle bulk). The other measures, occurring over the three lines drawn in at varying angulations, were termed error measurements.

Percent error was calculated using the following formula:

$$\text{Percent error} = \frac{\text{Difference in measurements (normal} - \text{error)}}{\text{Normal measurement}} * 100$$

3.B.5 Cross-sectional area measurement of TrA

Acquisition of ultrasound data:

Please refer to the methods section of Experiment Three for the detailed procedures regarding B-mode ultrasound data collection of TrA contraction.

Method of TrA measurement:

Using Adobe Photoshop 7.0, the muscular border of TrA from its posterolateral fascial insertion to a specified distance into the muscle bulk of TrA was outlined. This distance was standardized within each patient and represented the smallest linear distance from the

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fascial insertion into the muscle belly of TrA while keeping this line parallel to the fascial border of TrA. All cross-sectional area measurements were of muscle only, or in other words, between muscle-fascia boundaries (See Figure 12).[122] Image J [135] was then used to compute the cross-sectional area (CSA) of each image of TrA processed by Adobe Photoshop. (See Figure 13) Measurements of CSA were made in pixels^2 with each ultrasound image calibrated to the centimeter scale. From this information, cross-sectional areas were converted to units of centimeters^2.

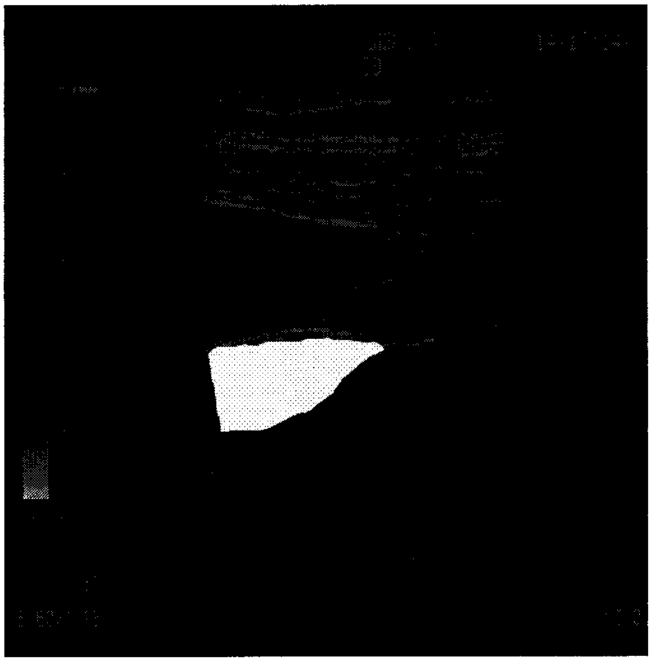

Figure 12: Adobe Photoshop image of the cross-sectional area of TrA (white) during a hollow contraction.

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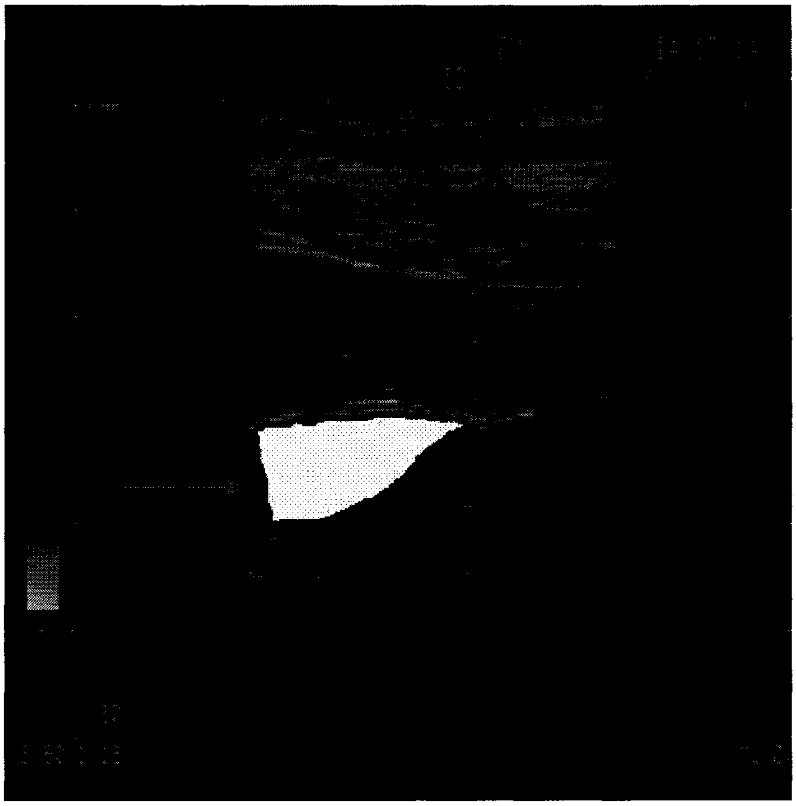

Figure 13: Representative outline (red) of the TrA muscle (red arrow) as performed by Image J software prior to CSA measurement.

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Method of TrA measurement specific to percent error

For percent error calculations, four thickness measurements were made (normal measurement and three error measurements). The error measurements were made at the location of the three additional lines (drawn in at varying angles) using Adobe Photoshop. Adobe Photoshop was also used to include the additional muscle slices that were created by the addition of lines bisecting the TrA muscle. Then, as described above, CSA measurements were made using Image J software. See Figure 14.

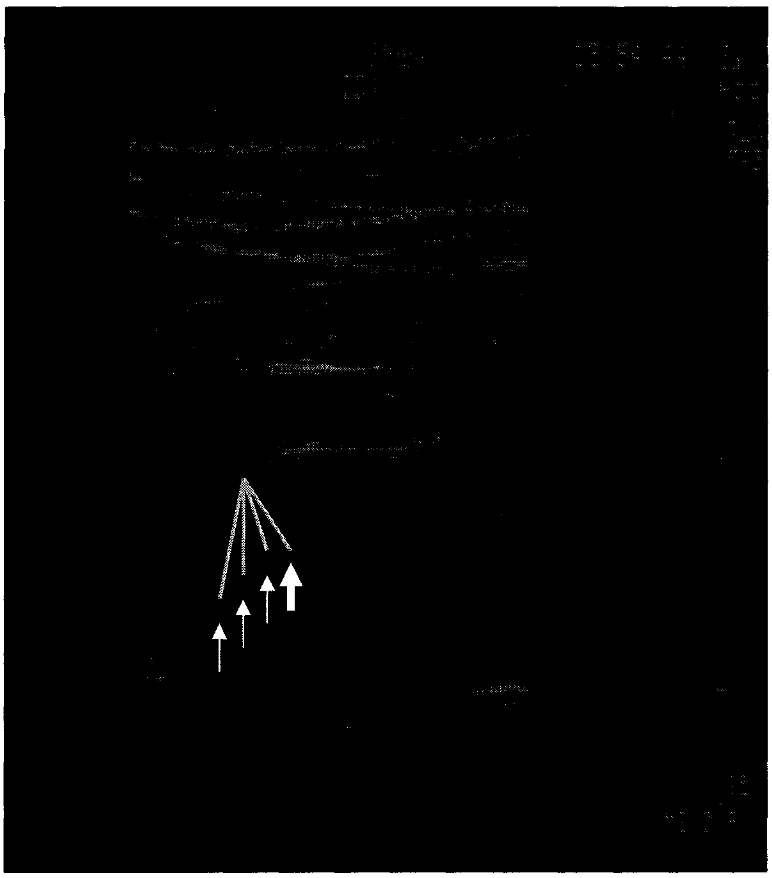

Figure 14. B-mode ultrasound image of TrA for percent error calculations. Normal thickness was measured at the line indicated by the thick white arrow and normal CSA was measured as the area from this line to the insertion of end of the TrA on the left. Thickness and CSA were then measured at the location of three additional lines added to ultrasound image (as indicated by the narrow white arrows) in order to calculate error. Not drawn to scale.

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3.B.6 Statistical analysis

Intra-class correlation coefficient (3,1) was used to determine the intra-rater reliability of this examiner.[134]

For the concurrent validity of measuring the posterior aspect of TrA using CSA, a one-sample t-test was used to determine if a significant difference was present between the TrA activation ratio calculated in this study to that already established in the literature as normal (≥ 2.0).

3.C Experiment Three: *The effect of two abdominal muscle contraction patterns on spinal stiffness.*

This experiment investigated the effect of two abdominal contraction patterns, abdominal hollowing and abdominal bracing, on stiffness in the lumbar spine as measured by an AI device. It was hypothesized that both contractions would increase spinal stiffness significantly compared to resting stiffness values. In addition, it was hypothesized that spinal stiffness generated by the hollow contraction would be significantly greater than changes in stiffness resulting from the brace contraction.

3.C.1 Subjects

Subjects were asymptomatic male and female volunteers. They were recruited primarily from 1) staff at the University of Alberta; 2) students attending the University of Alberta; and 3) friends/referrals of previous subjects.

3.C.2 Recruitment and consent

Subjects were recruited through advertisements placed throughout the University (See Appendix C). As well, recruitment e-mails were sent to the University of Alberta staff and students using departmental list serves (See Appendix D). Appropriate approval from

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the University of Alberta Health Research Ethics Board was received prior to the distribution of recruitment materials.

After a response from a potential subject was received, the researcher verbally informed him/her about the study and pre-screened for relevant inclusion and exclusion criteria. Following this contact, and if subjects wished to participate, a meeting time was arranged where subjects were given a project information letter (See Appendix E) and then provided with an opportunity to ask questions about the study. If still agreeable to participating in the study, subjects were given an informed consent form to read and sign (See Appendix F).

3.C.3 Sampling

A consecutive sampling method was used to ensure that sample size would be achieved. In this way, subjects continued to be recruited until the sample size was attained.

3.C.4 Sample size

It was calculated that a total of 28 subjects were required to yield a power of 80% toward detecting a difference in stiffness measures between abdominal hollowing and bracing when there truly is a difference. This was based on a significance level of 0.05. (See Appendix G for sample size calculations).

3.C.5 Inclusion criteria

The subjects who participated in this study were asymptomatic subjects between the ages of 18-50 without current low back pain or a history of low back pain within the last year. Fifty years of age was chosen as the upper limit of inclusion as it has been shown that at this age the prevalence of osteoporosis increases.[136] Osteoporosis has been identified as a contraindication to mobilization therapy,[42] a technique similar to AI. Lastly, to be included, subjects had to be able to lie flat in the prone position and be capable of performing the two stabilizing torso contractions to participate in this study.[59]

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3.C.6 Exclusion criteria

Subjects were excluded from this study primarily on the basis of back pain and/or medical conditions that could affect the safety of measurement of spinal stiffness using AI. Please refer to Table 2 in Experiment One for a full list of exclusion criteria. Also refer to Appendix E Project Information Letter for exclusion information.

3.C.7 Study design

This study utilized a design where subjects acted as their "own control". Each subject had the stiffness of his/her back measured at rest, during the hollow contraction, and during the brace contraction. This design permitted within-subject comparison of these three conditions.

The primary outcome measure was stiffness of the lumbar spine, as measured by AI. Secondary outcome measures included the measurement of muscle activity by surface electromyography (superficial trunk muscles) and CSA measurements of B-mode ultrasound images (TrA).

3.C.8 Data collection

All subject interviews and data collection occurred within the Common Spinal Disorders Research Lab (CH 3-44) in Corbett Hall at the University of Alberta.

3.C.9 Measurement of subject characteristics (See Appendix H):

The following items were obtained to help define the study population.

Body height

Body height was quantified using a body height meter. Subjects were measured barefoot. Subjects were instructed to stand with feet together and height was recorded in meters to two decimal places.

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Body mass

Subjects were measured barefoot on a digital scale wearing shorts (and a sports bra for female subjects). Mass was measured in kilograms to one decimal place.

Body mass index (BMI)

BMI was calculated using the following formula:

$$\text{BMI} = \frac{\text{mass in kg}}{(\text{height in m})^2}$$

3.C.10 Study procedure:

The study procedure was divided into three sessions: 1) the training session, 2) the transition to practice of contractions under experimental conditions, and 3) the testing session.

3.C.10.i Training session

Teaching of the abdominal contraction patterns:

Subjects were seen on an individual basis. Each subject was taught the abdominal hollowing contraction[34] and the abdominal brace contraction[65] using previously established descriptions of the contractions and teaching methodologies. Specifically, each contraction type was described to the subjects with emphasis on a low force of contraction (15-30% MVC).[34]

For the abdominal hollowing contraction, subjects were instructed to put their navel up and in towards the spine or to pull their lower abdomen away from the waist band of their pants. Tactile cues for the TrA (anterior and inferior to the anterior superior iliac spines and lateral to the rectus abdominis) and multifidus (muscle bellies adjacent to the lumbar spinous processes) were used to help facilitate contraction. The tactile cues consisted of instruction to gently swell out or contract their muscle against the researcher's fingers (particularly for multifidus). The contraction was demonstrated by the researcher then practiced by the subjects both in supine and prone positions to facilitate TrA and multifidus, respectively. Lastly, B-mode ultrasound imaging of the anterolateral

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abdominal wall was used in the teaching phase to facilitate proper contraction of TrA as this has been shown to increase the number of subjects able to properly perform the contraction and decrease the number of learning trials required.[127] Based on previous research, it was anticipated that approximately 10 trials would be needed to properly achieve a selective contraction of TrA[127] although in order to minimize muscular fatigue, the contractions during the learning trials were held for only 2 seconds. The researcher watched the subjects' performance of this contraction to identify improper substitution strategies such as aberrant movement, inappropriate contours in the abdominal wall, aberrant breathing patterns, and unwanted back extensor activity. See Appendix I for more detail on substitution strategies.

For the abdominal bracing contraction, the subjects began by standing and palpating their active low back extensors while the lumbar torso was slightly flexed. They were instructed to slowly extend (straighten out) until they felt their back extensors "shut off" (could no longer feel a swelling of the muscle under their fingers). This position was considered the position of rest for the spine. The subjects were instructed that without moving from this position, they should contract the abdominal muscles and feel the extensors contract once again. This isometric activity in both trunk flexors and extensors was considered the abdominal brace.[65] Tactile cues were used globally over the abdominal and back musculature to help promote contraction in these areas, while avoiding a draw-in maneuver. Again, a demonstration of this contraction was performed by the researcher. Similar methodology regarding number of practice trials and length of holding time of the brace contraction were used as discussed above in the abdominal hollowing section.

Verification of abdominal muscle pattern:

In order to differentiate between the two contractions (hollow and brace), B-mode ultrasound and surface EMG were used concurrently to determine the activation of both the deep and superficial trunk muscles, respectively. These technologies were used both in training sessions and during data collection. Discussed below are the features of the

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different contractions that were required to be present before any contraction could be considered as acceptable.

For verification of proper abdominal hollowing, specific contraction of the TrA was required via drawing in the abdominal wall. Real-time ultrasound imaging of the posterolateral abdominal wall at rest and during the abdominal hollowing contraction was used to demonstrate a contraction of the TrA. Images of both the relaxed state and the contracted state (taken at the same time as stiffness measures) were used to verify the change in muscle thickness/CSA occurring during this specific abdominal contraction (increased thickness of TrA with contraction).[59] Surface EMG of the obliques, rectus abdominis, and the erector spinae muscles was used to determine the contraction level of these global muscles. Contraction of the global muscles was not expected to occur with the performance of the hollowing contraction.[59] In summary, the hollow contraction should create increased TrA size, confirmed by ultrasonic imaging, with minimal to no EMG signal from other muscles. (See Table 3)

For verification of proper abdominal bracing, a general contraction of all the abdominal and low back muscles was required while the subject maintained a neutral spine. Real-time ultrasound imaging and collection of surface EMG was performed while the subject was at rest (relaxed abdominal wall) and during a brace contraction. For the abdominal brace, an increase in muscle size of all muscles of the posterolateral abdomen was expected as a global contraction is being performed. More importantly, surface EMG activity of the obliques, rectus abdominis, and the erector spinae muscles should exhibit higher values than at rest or during the abdominal hollowing contraction.[59] (See Table 3).

	Rest			Hollow			Brace		
	TrA (CSA)	***RA, EO, IO*** (EMG)	***T-ES, L-ES*** (EMG)	***TrA*** (CSA)	***RA, EO, IO*** (EMG)	***T-ES, L-ES*** (EMG)	***TrA*** (CSA)	***RA, EO, IO*** (EMG)	***T-ES, L-ES*** (EMG)
Expected	None	None	None	↑	Min.	Min.	↑	↑	↑

Table 3: Expected contraction of the trunk muscles for each condition: rest, hollow, and brace. TrA – transversus abdominis; RA- Rectus Abdominis; EO – External Obliques; IO – Internal Obliques; T-ES – Thoracic Erector Spinae; L-ES – Lumbar Erector Spinae; Min. – minimum contraction.

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3.C.10.ii. Transition to practice of contractions in experimental conditions:

Prior to beginning the formal testing procedure and in addition to the training session for the contractions, subjects performed the two abdominal contractions while formally recording with sEMG and B-mode ultrasound so that the muscle patterns could be verified and the equipment's recording ability could be confirmed. Then, the hollow and the brace contraction were practiced adding a preliminary indentation (not included in formal testing procedure) to ensure that the proper contraction was being performed and maintained during AI. If performing the contractions improperly when indentation was added, the subjects were given feedback on the necessary muscular contraction alterations needed to achieve the proper contraction. The muscle contraction (with stiffness testing) was then re-attempted. In the case of the abdominal brace, subjects were instructed to lift their heads off the bed in order to elicit erector spinae contraction (which was occasionally inhibited by the presence of the indentor on the L4 spinous process).

3.C.10.iii Testing session

Testing protocol:

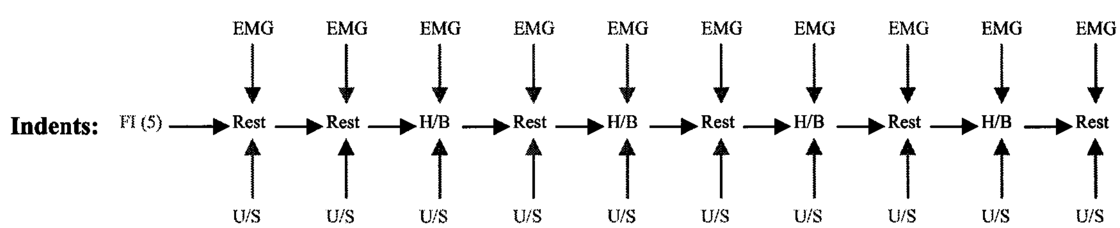

Figure 15: Testing protocol. Indents refer to stiffness measurements. FI refers to the five familiarization indentations performed prior to data collection. H/B refers to the hollow or brace contraction (randomly chosen). H/B refers to the hollow or brace contraction opposite to the previously performed contraction. U/S refers to B-mode ultrasound measurement; two images were taken during each indent. EMG refers to surface electromyography.

With the subject lying in a prone position, spinal stiffness was measured at the L4 vertebral level by AI. Indentation testing was performed by one researcher (TL) who had logged over 150 hours of experience operating the assisted indentor.

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Following 5 familiarization trials, a total 10 indentations were performed (See Figure 15) during which the subject was at rest, or performed either the abdominal hollow and abdominal brace a total of two times (6 measurements during rest, 2 measurements during the hollow contraction, and 2 measurements during the brace contraction). Real-time B-mode ultrasound of the TrA (two images taken) and sEMG of the superficial trunk muscles were recorded for each indentation trial, including the stiffness testing with the subjects at rest. A standard rest period of two minutes in the prone position was used between each indentation to ensure that the subject fully relaxed between measurements. Visual inspection of the limb musculature was performed to ensure that there were no signs of increased muscle activity. The order of testing for the effect of the abdominal hollowing and the abdominal brace was randomized.[59] The number of abdominal contractions performed by the patients in this study was similar to that previously used in the literature, thus it was anticipated that, in combination with rest periods, fatigue of the abdominal muscles would not occur.[59]

3.C.11. Measurement of spinal stiffness using an assisted indentation device:

Please refer to the methods section of Experiment One for details pertaining to the assisted indentor instrument specification, protocol of spinal stiffness measurement, and measurement of spinal stiffness.

3.C.12. Measurement of superficial muscle activity using surface electromyography:

Instrument:

Surface electromyography (EMG) of the superficial trunk muscles was performed using the AMT-8 system (Bortec Biomedical).

Procedure:

The subject's skin was shaved over the EMG sites and cleaned with an alcohol swab in order to reduce impedance.[106] After allowing the skin to dry, Ag/AgCl bipolar disposable electrodes (Bortec BiPole^{TM}) with an active diameter of 1 cm, and an inter-electrode distance of approximately 2 cm were placed on the skin. Five channels of EMG were collected from the right side of the lumbar spine: rectus abdominis (3 cm lateral to the

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umbilicus), external oblique (approximately 15 cm lateral to the umbilicus), internal oblique (approximately midway between the anterior superior iliac spine and symphysis pubis, above the inguinal ligament), thoracic erector spinae (5 cm lateral to T9 spinous process), and lumbar erector spinae (3 cm lateral to L3 spinous process)[22, 137] with a reference electrode placed over the clavicle. (See Figure 16) To reduce the occurrence of cross-talk, specific manual muscle tests were performed for each muscle while the researcher viewed the raw EMG output. Criteria for the absence of cross-talk (and the ability to continue on with the study protocol) included an increase in raw EMG signal for the appropriate muscle being tested, with an absence of signal increase in surrounding muscles.

Figure 16: Electrode placement for the abdominal muscles (picture to the left) and the erector spinae muscles (picture to the right).

A series of maximal contractions against resistance were undertaken for normalization of the EMG magnitude. For the abdominal muscles, each subject was situated in a sit up position and manually braced by the researcher. A maximal isometric flexor moment followed sequentially by a right and left twist moment was performed.[137] For the extensor muscles, a resisted maximal extension was performed with the subjects in prone.[14].

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Analysis of surface EMG signals:

The raw sEMG signals were then A/D converted with a 16-bit, 16 channel converter at 2,000 Hz, full wave rectified and low pass filtered with a second order single pass Butterworth filter.[137] A cut-off frequency of 2.5 Hz was used.[138] Further analysis of the sEMG signal was performed using customized Labview software. Average EMG amplitude over 1000 ms was taken at the trigger points where ultrasound images were collected in time (detailed below). The filtered EMG data was then normalized to MVC amplitudes (EMG baseline was subtracted from the EMG magnitude during contraction, which was then normalized as a proportion of the maximum voluntary contraction EMG magnitude). Analysis of the amplitude of the MVC EMG signals was performed; no frequency analysis was completed.

3.C.13. Measurement of TrA muscle activity using B-mode ultrasound

Instrument:

Real-time ultrasound imaging of the posterolateral abdominal wall was performed using a Sonoline Sienna Siemens B-mode ultrasound (Siemens Medical Systems, Inc.; Issaquah, WA) with a 7.5-mHz linear array transducer. This was performed by 2 separate volunteers who were trained in ultrasound acquisition prior to the onset of this study.

Procedure:

The gelled transducer was positioned 25 mm postero-medial to the midpoint between the ribs and ilium on the mid-axillary line and parallel to the muscle fibers of TrA. To standardize the location of the transducer, two methods were utilized. First, the location of the transducer was marked so that identical placement would be used for all measurements.[120] (See Figure 17) Second, to ensure the imaging location remained consistent, the hyperechoic interface between the TrA and thoracolumbar fascia was positioned in the far right side of the ultrasound image.[123] (See Figure 17) The angle of the transducer was then adjusted to optimize visualization of the image.[123] If needed, the image gain and focus were adjusted to produce the clearest picture of the tissues. A foot pedal switch was used to trigger image acquisition and this same pedal marked the EMG

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record at the time when the images were taken. As a result, EMG and ultrasound imaging were synchronized. Images were transmitted to a customized labview program at a resolution of 640 X 480 pixels. To reduce the confounding effect of transducer movement on the subjects' skin that may occur between indentations, two pictures of the B-mode ultrasound images were taken for each contraction/indentation: 1) at rest before contraction and, 2) during the maximal indentation portion of the stiffness measurement while the subject held the contraction. (See Figures 18 and 19)

Figure 17: Pen marking of the location of the ultrasound transducer during testing.

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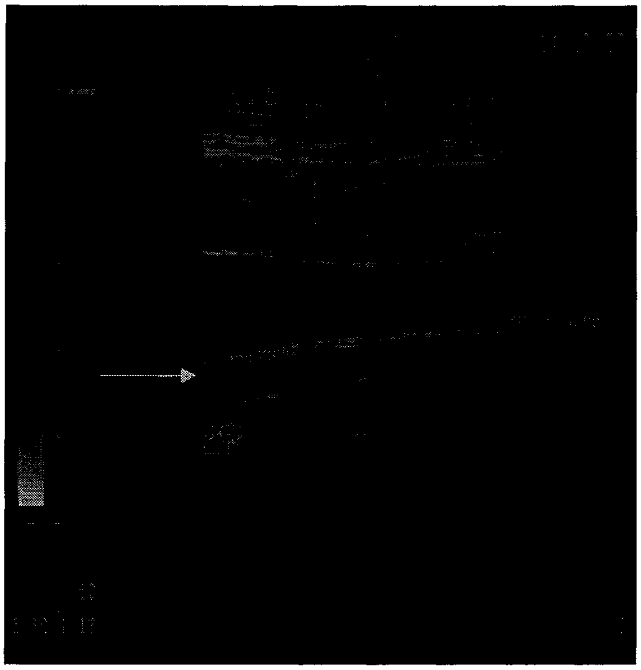

Figure 18: Ultrasound image of the TrA (indicated by the yellow arrow) with the subject at rest. Note the position of the hyperechoic interface between the TrA and thoracodorsal fascia (indicated by thick red arrow) at the far right of the ultrasound image.

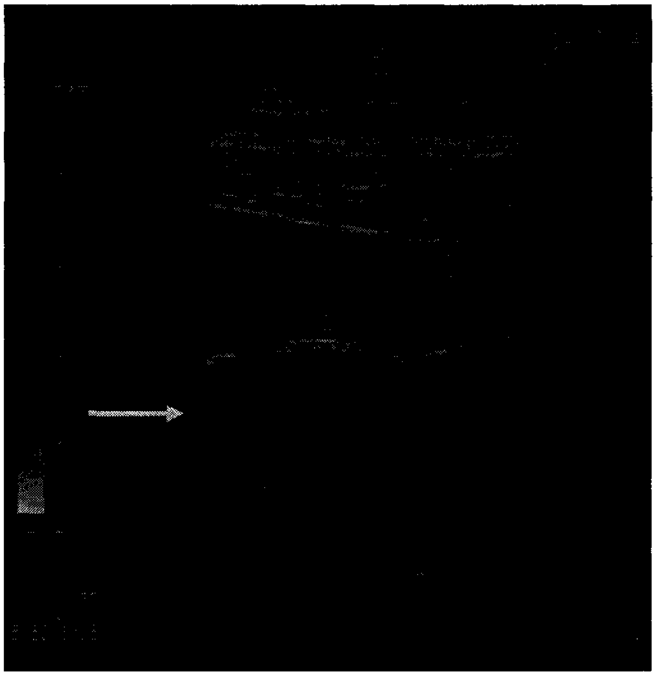

Figure 19: Ultrasound image of TrA (indicated by the yellow arrow), while contracting the TrA (hollow contraction).

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In the literature, the anterior aspect of TrA is measured exclusively to determine TrA contraction.[59, 118, 121-124, 127] However, in this study, because indentation occurred simultaneously with ultrasound, the subjects were required to be prone. In this way, the anterolateral aspect of the abdomen could not be accessed. Instead, the posterolateral aspect of TrA and its insertion into the TLF was imaged.

Analysis of B-mode ultrasound data

Please refer to the methods section of Experiment Two for detailed information pertaining to the analysis of ultrasonic data. In Experiment Three, in addition to the calculation of CSA, the mean thickness of TrA was calculated by dividing the CSA by the length over which the CSA was taken. This allowed comparison to other studies that used thickness measures of TrA contraction.

3.C.14. Statistical analysis

Descriptive statistics (mean, standard deviation) were reported for all subject demographic data (age, weight, height, and body mass index). As well, the mean and standard deviation for spinal stiffness, superficial trunk muscle activity, and TrA contraction values during the three conditions of contraction: rest, hollow, and brace were reported.

As reported previously in Experiment One, for data analysis purposes, all five of the familiarization trails in Experiment Three were discarded. In addition, the first trial (stiffness measurement during rest) of the ten experimental indentations was discarded as this first trial has been shown to highly variable.[45, 93]

For ease of analysis, a repeated measures ANOVA was used to determine if stiffness data from the four indentation trials for the rest condition could be pooled. If no significant differences were found, data from the four rest conditions were averaged and this value used for further analysis. Additionally, paired t-tests were used to determine if the stiffness data from the two indentation trials for the two muscle contractions (hollow and brace condition) could be pooled. Again, if no significant differences were found the

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average of the two contractions for the conditions of the hollow and the brace was taken for further analysis. This pooling analysis was also performed for EMG and ultrasound data in the same manner.

Following pooling, a repeated measures ANOVA was used to determine if significant differences occurred in the stiffness values during the rest, hollow, and brace conditions. This analysis was repeated for both EMG data and ultrasound data to determine if significant differences in superficial muscle activity and TrA contraction, respectively, occurred during the rest, hollow, and brace conditions.

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CHAPTER FOUR
RESULTS

Overview:

In Experiment One, excellent reliability was demonstrated for the stiffness outcome variables, GS and MMS. These variables displayed ICC values of 0.91 and 0.93, respectively. According to Fleiss, [130] an ICC value above 0.75 is indicative of excellent reliability. In support of this finding, the inter-trial inconsistency (ITI) of AI was found to be less than 10%. Specifically, GS had an ITI of 6.23 % (+/- 4.52%) and MMS demonstrated an ITI of 7.71% (+/- 5.33%). With respect to assessing differences in stiffness values over time, Pearson's r demonstrated significant correlations between all indentations (2-10) for both measurements of stiffness (GS and MMS). Further, paired t-tests showed a significant difference between the second and last indentation for GS values ($p = 0.00$) while this was not observed for MMS values ($p = 0.82$).

In Experiment Two, excellent reliability was also exhibited for the measurement of TrA CSA with the ICC value found to be 0.998. In addition, the technique of measuring TrA CSA using a posterior view was found to be valid as one sample t-tests did not demonstrate a significant difference between the TrA activation ratio calculated in this study and previously calculated values ($p = 0.22$, males; $p = 0.51$, females).

For Experiment Three, significant differences were present between the stiffness measured during rest, hollow contraction, and brace contraction. Specifically, stiffness of the 4^{th} lumbar vertebrae was significantly greater during the hollow and brace contractions than when measured at rest. Additionally, the brace contraction generated greater stiffness at L4 than the hollow contraction. (See Table 4) Surface electromyography values demonstrated a significant difference between all contractions for all superficial trunk muscles, with the exception of the erector spinae muscles, for which no significant difference was present between rest and the hollow contraction. Lastly, B-mode ultrasound values of TrA muscle contraction exhibited significant differences for comparisons involving the rest condition. No differences in TrA muscle

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contraction occurred between the hollow and brace conditions. (See Table 4) See Table 5 for the observed contraction of trunk muscles for each condition.

Contraction Comparisons	Stiffness		EMG		U/S	
	GS	MMS	Abs	ES	CSA	Mean Thickness
Rest vs. Hollow	**	**	**		**	**
Rest vs. Brace	**	**	**	**	**	**
Hollow vs. Brace	*	*	**	**		

Table 4: Results of statistical comparisons between the rest, hollow, and brace conditions for stiffness (N/mm), EMG (% of MVC), and ultrasound values of transversus abdominis contraction (CSA in cm^2, mean thickness in cm). GS = global stiffness, MMS = mean maximal stiffness, Abs = abdominals (rectus abdominis, external obliques, and internal obliques), ES = erector spinae muscles, U/S = ultrasound, CSA = cross-sectional area. *denotes significance at the $p < 0.05$ level. **denotes significance at the $p < 0.01$ level. Exception: Rectus abdominis had a significance of $p < 0.05$ for the rest-hollow comparison.

	Rest			Hollow			Brace		
	TrA (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)	*TrA* (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)	*TrA* (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)
Observed	None	None	None	↑	Min.	None	↑	↑	↑

Table 5: Observed contraction of the trunk muscles for each condition: rest, hollow, and brace. TrA – transversus abdominis; RA – Rectus Abdominis; EO – External Obliques; IO – Internal Obliques; T-ES – Thoracic Erector Spinae; L-ES – Lumbar Erector Spinae; Min. – minimum contraction.

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4.A Experiment One

4.A.1 Subject demographics

A total of thirty subjects were recruited to participate in this project with three excluded due to previous back or lower extremity injury within the last year, two excluded for exceeding the age limit, and two excluded due to intolerance of the indentation procedure. Of the two subjects excluded for intolerance of the indentation procedure, one subject reported low back discomfort following indentation (excluded due to intolerance of the indentation procedure), although this was self-limiting and resolved within approximately one week and the other subject did not report lasting discomfort. This resulted in 12 male and 11 female subjects who participated in this study (n = 23). (See Table 6 for subject demographics)

	Male (n = 12)	Female (n = 11)
Age (years)	26.17 (+/- 3.10)	24.45 (+/- 3.21)
Height (m)	1.79 (+/- 0.065)	1.63 (+/- 0.052)
Weight (kg)	76.23 (+/- 9.64)	58.41 (+/- 8.28)
Body Mass Index (kg/m^2)	23.85 (+/- 2.39)	21.59 (+/- 2.21)

Table 6: Mean (+/- standard deviation) of subject demographic characteristics.

4.A.2 Reliability and inter-trial inconsistency results

In this experiment, the reliability of the stiffness measures was described by the ICC which was calculated to be 0.91 for GS and 0.93 for MMS. Additionally, an estimate of the consistency in stiffness measures was obtained by calculating the inter-trial inconsistency which was 6.23% (+/- 4.52%) for the GS and 7.71% (+/- 5.33%) for MMS. (See Figure 20 and Figure 21 for individual subject representation of ITI values)

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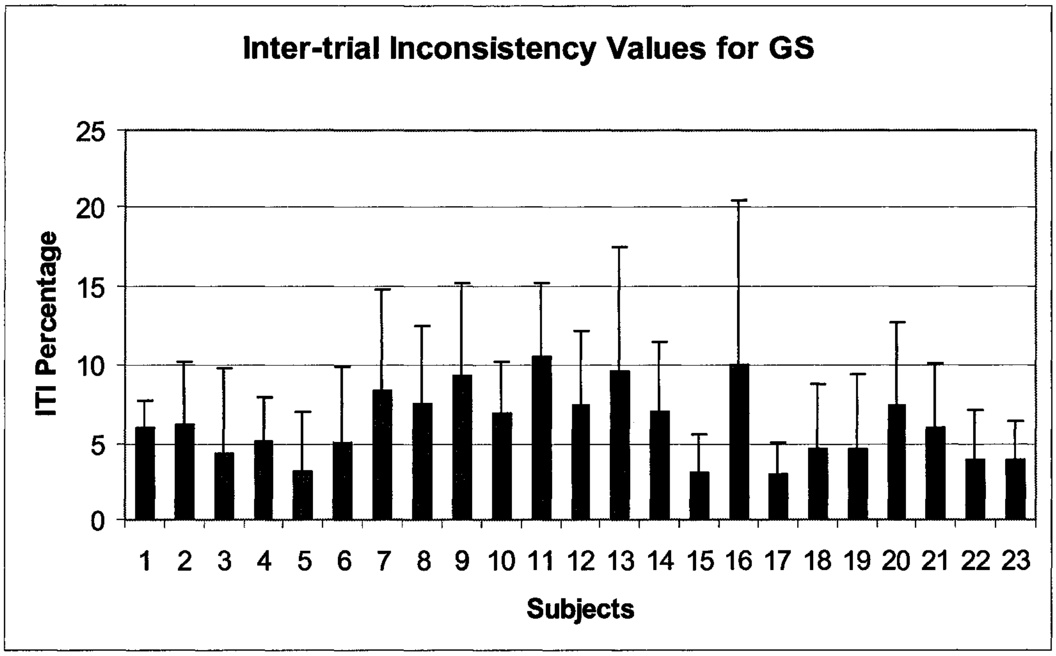

Figure 20: Inter-trial inconsistency values (+/- standard deviation) for GS estimates of L4 stiffness values.

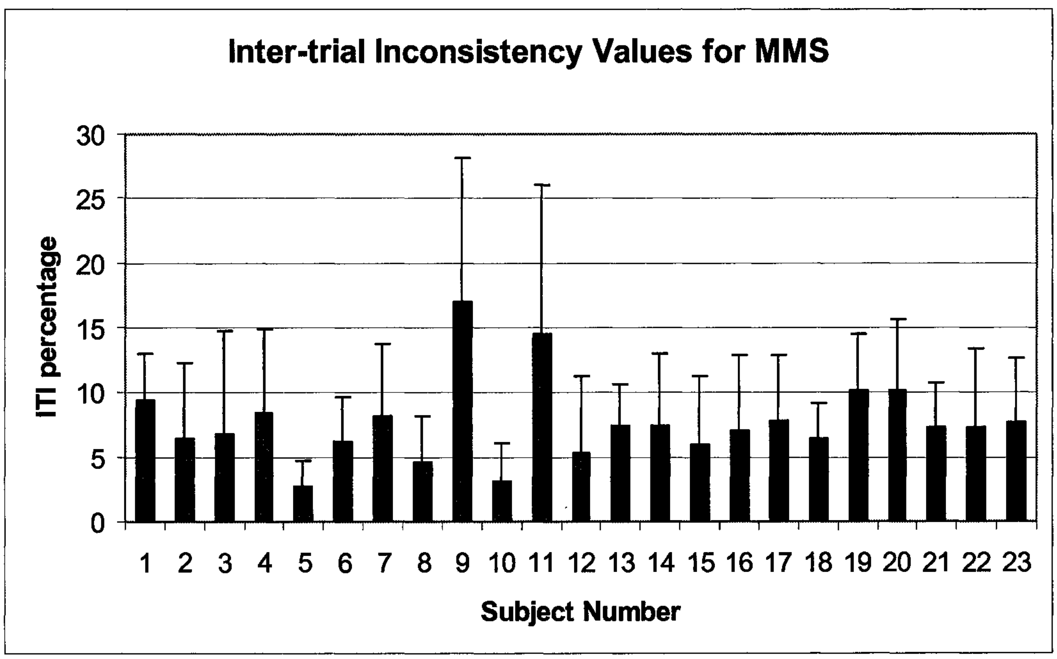

Figure 21: Inter-trial inconsistency values (+/- standard deviation) for MMS estimates of L4 stiffness values

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4.A.3 Measurement of changes in stiffness data as function of time

Pearson's r found significant correlations between all indentations (2-10) for both GS and MMS ($p < 0.01$). Specifically, Pearson's r values ranged from 0.84 – 0.96 for GS and from 0.91 – 0.96.

However, a paired t-test (2-tailed) found a significant difference between trial 2 and 10 for stiffness measurements of GS ($p = 0.00$) and no significant difference for MMS ($p = 0.82$). See Figure 22 and Figure 23 for the graphical representation of the change in stiffness values over time.

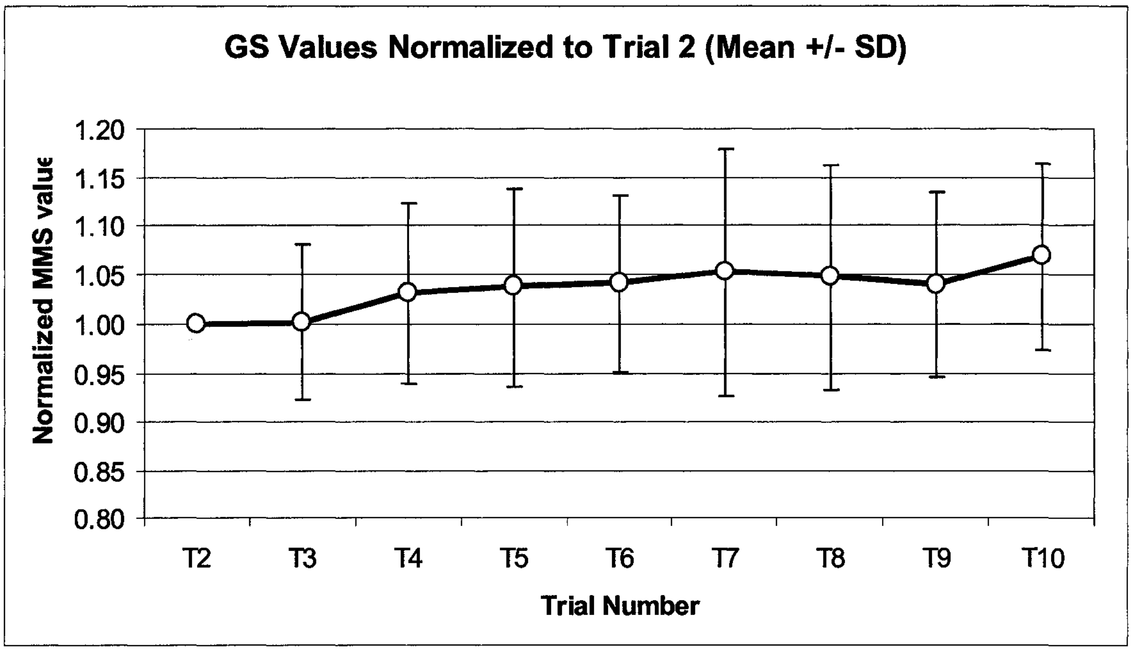

Figure 22: Change in GS stiffness values over time for all subjects.

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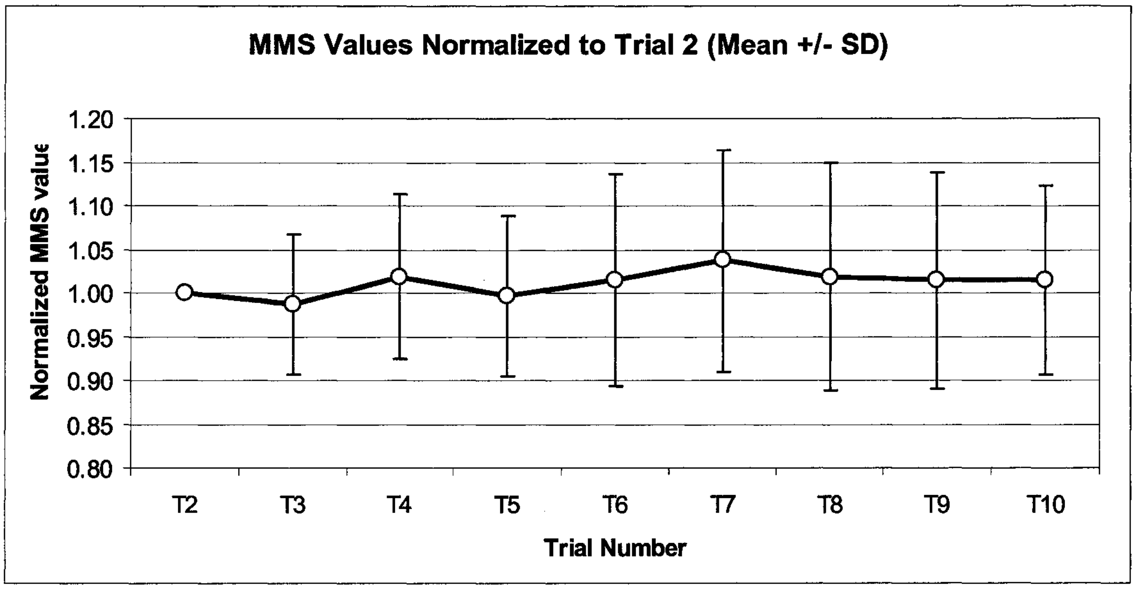

Figure 23: Change in MMS stiffness values over time for all subjects.

4.B Experiment Two

4.B.1 Subject demographics

Regarding the reliability component of this experiment, images from ten subjects chosen randomly from Experiment Three were used in this study. As this is a sub-population of Experiment Three, the demographics of the subjects whose images were used in this study are presented in the following Table (See Table 7).

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	Male	Female
Gender Number	5	5
Age (years)	27.6 (7.33)	24.4 (3.65)
Height (m)	1.78 (0.08)	1.67 (0.06)
Weight (kg)	76.04 (11.76)	60.29 (11.44)
BMI (kg/m^2)	23.92 (2.99)	21.42 (2.52)
Contractions Used	1 – Brace 3 – Hollow 1 – Rest	3 – Brace 2 - Rest

Table 7: Subject demographics for the ten subjects chosen randomly from Experiment Three; data used in the reliability study of CSA measurement of TrA.

Regarding the concurrent validity component of this experiment, data for the 28 subjects that participated in Experiment Three was used. A total of 224 ultrasound images were used (112 ultrasound image pairs). See Table 9 in the results section of Experiment Three for subject demographics.

4.B.2 Reliability and validity results

To measure intra-rater reliability, the intra-class correlation coefficient (3,1) was calculated and found to be 0.998 for CSA values measured during TrA contraction on two separate occasions.

The TrA activation ratio was calculated to be 1.83 for males and 1.89 for females. Further, results from a one sample t-test did not demonstrate a significant difference between the activation ratios calculated in this study and previously published activation ratios for TrA contraction taken from an anterior view (≥ 2.00)[123] (See Table 8 for one-sample t-test results).

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	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
Male activation ratio vs. 2.00	-0.17	0.13	-0.44	0.10	1.33	0.21
Female activation ratio vs. 2.00	-0.11	0.13	-0.39	0.17	0.88	0.39

Table 8: One-sample t-test for TrA activation ratios; male and female results. Calculated TrA activation ratio was compared to that previously reported in the literature (= 2.00).[123]

4.B.3. Percent Error

In all error situations, the CSA measurement technique demonstrated less error than the thickness measurement technique using a posterolateral image of TrA. See Table 9 for percent error values.

	Normal Measurement	Error Measurement	Difference Value	Percent Error
4 degrees off				
CSA	3.25	3.36	0.11	3.23%
Thickness	1.49	1.57	0.08	5.37%
13.5 degrees off				
CSA	3.25	3.56	0.31	9.50%
Thickness	1.49	1.66	0.17	11.41%
21.1 degrees off				
CSA	3.25	3.85	0.59	18.27%
Thickness	1.49	1.81	0.32	21.48%

Table 9: Percent error values for TrA measurements of CSA and Thickness for three error situations. CSA measured in cm^2 and thickness measured in cm.

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4.C Experiment Three

4.C.1 Subject demographics

A total of 37 subjects were approached to participate in this study. Prior to any data collection, one subject was excluded from participation due to age (over the 30 year limit), two subjects due to lower extremity injury within the last year, one subject due to known spondylolisthesis, and one subject due to scheduling conflicts. This resulted in a total of 32 subjects completing the study protocol although 5 further subjects were excluded due to inability to perform the abdominal brace contraction (n = 2), inability to perform the abdominal hollow contraction (n = 1), inability to tolerate indentation (n = 1), and due to scoliosis that was not detected from the intake questionnaire (n = 1). Therefore, a total of 28 subjects' data (14 male and 14 female) was used for statistical analysis in this study. See Table 10 for subject demographics.

Male		**Mean**	**SD**	**Range**
	Age (years)	28.1	7.42	19 – 47
	Weight (kg)	77.11	8.44	67 – 91
	Height (m)	1.79	0.0065	1.71 – 1.92
	BMI (kg/m^2)	23.96	1.90	20.85 – 28.21
Female				
	Age (years)	26.5	5.49	20 – 41
	Weight (kg)	59.86	9.31	27 – 76.5
	Height (m)	1.66	0.0052	1.58 – 1.71
	BMI (kg/m2)	21.61	2.49	18.69 – 27.60

Table 10: Subject demographic characteristics.

4.C.2 Pooling of variables

If possible, the decision was made to pool similar data for ease of analysis. In this study, pooling was considered on the basis of experimental condition (rest, hollow, and brace) only. Gender was not considered as a pooling factor due to previous studies documenting gender differences in spinal stiffness, [96] trunk muscle responses as measured by EMG, [139] and ultrasound imaging of TrA.[140]

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4.C.2.i Pooling of stiffness values

The results of the repeated measures ANOVA for rest conditions demonstrated that for both GS and MMS values, no significant differences were present (p = 0.98 and 0.22). Similarly, paired t-test results demonstrated that for both GS and MMS values, no significant differences were present between hollow contraction 1 and hollow contraction 2 or between brace contraction 1 and brace contraction 2 (p = 0.20 – 0.75). Therefore, these values were pooled for these conditions for further analyses. (See Table 11) Please refer to Appendix J and K for the repeated measures ANOVA and paired t-test results for GS values and MMS values, respectively. For GS and MMS values ***prior*** to pooling, see Appendix L.

	Contraction	Mean	SD	Range of Averaged Data
GS				
Male	Rest	9.04	0.83	7.67 – 10.36
	Hollow	10.54	1.61	7.63 – 13.52
	Brace	12.31	4.82	7.16 – 27.03
Female	Rest	7.95	1.46	5.78 – 10.06
	Hollow	9.91	2.16	5.89 – 13.73
	Brace	11.44	3.81	5.50 – 21.10
MMS				
Male	Rest	7.17	1.25	5.33 – 9.44
	Hollow	8.78	1.95	6.33 – 12.49
	Brace	10.35	3.19	6.41 – 17.95
Female	Rest	6.92	1.35	4.48 – 9.06
	Hollow	8.94	1.91	4.52 – 11.96
	Brace	9.84	3.08	4.73 – 16.97

Table 11: Average stiffness values following data pooling; both GS and MMS measured in N/mm.

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4.C.2.ii Pooling of EMG values

No significant differences were demonstrated for rest conditions for any of the trunk muscles as shown by the results of the repeated measures ANOVA ($p = 0.18 - 0.65$). Further, paired t-tests exhibited no significant differences between the two hollow contractions nor between the two brace contractions for EMG values for any of the muscles (0.12 – 0.95). Therefore, for further analyses, EMG values were pooled for each condition (rest, hollow, and brace), for each muscle. (See Tables 12 - 16 for muscle specific average EMG values). Please refer to Appendix M for the repeated measures ANOVA and paired t-test results for EMG values. For EMG values ***prior*** to pooling, see Appendix N.

RA	Contraction	Mean	SD	Range of Averaged Data
Male				
	Rest	0.07	0.18	0.00 – 0.58
	Hollow	0.38	0.56	0.00 – 1.61
	Brace	2.78	3.18	0.00 – 9.85
Female				
	Rest	0.08	0.29	0.00 – 0.11
	Hollow	0.69	1.14	0.00 – 2.63
	Brace	1.93	2.32	0.00 – 8.89

Table 12: Normalized average EMG values for Rectus Abdominis during Rest, Hollow, and Brace Conditions following data pooling. EMG activity (Volts) expressed as a % of MVC.

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EO	Contraction	Mean	SD	Range of Averaged Data
Male				
	Rest	0.03	0.09	0.00 – 0.24
	Hollow	2.36	2.02	0.00 – 6.25
	Brace	18.47	14.23	4.62 – 50.94
Female				
	Rest	0.64	1.67	0.00 – 6.25
	Hollow	2.80	2.90	0.00 – 8.82
	Brace	19.49	9.53	8.68 – 42.71

Table 13: Normalized average EMG values for External Obliques during Rest, Hollow and Brace Conditions following data pooling. EMG activity (Volts) expressed as a % of MVC.

IO	Contraction	Mean	SD	Range of Averaged Data
Male				
	Rest	0.13	0.27	0.00 – 0.70
	Hollow	11.01	16.01	0.00 – 61.11
	Brace	35.28	20.43	6.00 – 64.47
Female				
	Rest	0.36	0.89	0.00 – 3.13
	Hollow	15.90	12.46	0.00 – 31.03
	Brace	34.23	23.06	5.14 – 73.53

Table 14: Normalized average EMG values for Internal Obliques during Rest, Hollow, and Brace Conditions following data pooling. EMG activity (Volts) expressed as a % of MVC.

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T-ES	Contraction	Mean	SD	Range of Averaged Data
Male				
	Rest	0.11	0.23	0.00 – 0.59
	Hollow	0.15	0.33	0.00 – 1.16
	Brace	7.83	7.14	0.00 – 20.93
Female				
	Rest	0.43	1.32	0.00 – 0.46
	Hollow	0.69	1.60	0.00 – 5.17
	Brace	12.94	13.66	0.00 – 44.83

Table 15: Normalized average EMG values for Thoracic Erector Spinae during Rest, Hollow, and Brace Conditions following data pooling. EMG activity (Volts) expressed as a % of MVC.

L-ES	Contraction	Mean	SD	Range of Averaged Data
Male				
	Rest	1.02	0.86	0.00 – 2.33
	Hollow	0.43	0.83	0.00 – 2.94
	Brace	20.09	15.48	1.87 – 50.00
Female				
	Rest	0.89	0.93	0.00 – 3.65
	Hollow	0.60	0.87	0.00 – 2.63
	Brace	20.54	17.35	1.89 – 63.16

Table 16: Normalized average EMG values for Lumbar Erector Spinae during Rest, Hollow and Brace Conditions following data pooling. EMG activity (Volts) expressed as a % of MVC.

4.C.2.iii Pooling of ultrasound values

The results of the repeated measures ANOVA demonstrated that for TrA CSA and TrA mean thickness values for rest conditions, no significant differences were present ($p = 0.57$ and 0.45, respectively). Similarly, paired t-tests did not find any significant differences between the two hollow contractions nor between the two brace contractions for either TrA CSA or mean thickness ($p = 0.46 – 0.89$). Further analyses use the average TrA CSA and TrA mean thickness values for each condition (rest, hollow, brace). (See

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Table 17) Please refer to Appendix O and P for the repeated measures ANOVA and paired t-test results for TrA CSA and TrA mean thickness measurements, respectively. For TrA CSA and mean thickness values ***prior*** to pooling, see Appendix Q.

	Contraction	Mean Difference	SD	Range
TrA CSA				
Male				
	Rest	0.18	0.26	-0.36 – 1.00
	Hollow	1.85	1.09	0.45 – 4.31
	Brace	2.63	1.44	0.51 – 5.84
Female				
	Rest	0.07	0.31	-0.60 – 0.70
	Hollow	1.79	0.70	0.44 – 3.03
	Brace	1.86	1.19	0.34 – 4.90
TrA Mean Thickness				
Male				
	Rest	0.03	0.04	-0.03 – 0.10
	Hollow	0.45	0.25	0.21 – 1.02
	Brace	0.62	0.27	0.23 – 1.04
Female				
	Rest	0.02	0.04	-0.06 – 0.08
	Hollow	0.44	0.15	0.25 – 0.75
	Brace	0.44	0.21	0.11 – 0.94

Table 17: Descriptive statistics for average TrA CSA values and mean thickness values for ultrasound during the rest, hollow, and brace conditions (measured in cm^2 and cm, respectively) following data pooling. Mean difference = TrA (CSA or mean thickness) Time 2 – TrA (CSA or mean thickness) Time 1.

4.C.3 Analysis of the average stiffness values (GS and MMS) for rest, hollow, and brace conditions

Global stiffness values were found to demonstrate a significant effect for the experimental condition (rest, hollow, brace). (See Table 18) More specifically, pair-wise comparisons exhibited significant differences between all three conditions. It was found

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that both the hollow and the brace produced significantly higher stiffness values than those measured during rest ($p < 0.00$), while the brace condition demonstrated significantly higher stiffness values than the hollow condition ($p = 0.02$). (See Table 19 and Figure 24) Further, no gender differences were found for GS stiffness values ($p = 0.28$) for any of the three conditions and the covariate of age was not significant ($p = 0.549$). However, the covariate of BMI was significant ($p < 0.05$) for GS values. Specifically, a positive relationship was seen between BMI and stiffness for all conditions ($r^2 = 0.02$, 0.03, and 0.17 for the rest, hollow, and brace condition, respectively). Lastly, the increase in GS values during the hollow contraction (as compared to resting GS values) was calculated to be 20.6% and during the brace contraction was calculated to be 40.05%. See Table 20 for gender-specific percentages.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Condition	160.34	1.26	127.19	14.03	0.00*
Condition X Gender	0.73	1.26	0.58	0.06	0.86
Error (exercise)	297.12	32.78	0.07		

Table 18: Repeated Measures ANOVA for GS values for rest, hollow, and brace conditions; male and female. GS measured in N/mm. Sphericity not assumed; Huynh-Feldt adjustment used.

(I) Condition	(J) Condition	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-1.73*	0.31	0.00	-2.36	-1.10
	Brace	-3.38*	0.82	0.00	-5.07	-1.70
Hollow	Rest	1.73*	0.31	0.00	1.10	2.36
	Brace	-1.66*	0.68	0.02	-3.05	-0.27
Brace	Rest	3.38*	0.82	0.00	1.70	5.07
	Hollow	1.66*	0.68	0.02	0.27	3.05

Table 19: Pair-wise comparison between GS values for the rest, hollow, and brace conditions. GS measured in N/mm. Based on estimated marginal means. * Mean difference is significant at the p < 0.05 level.

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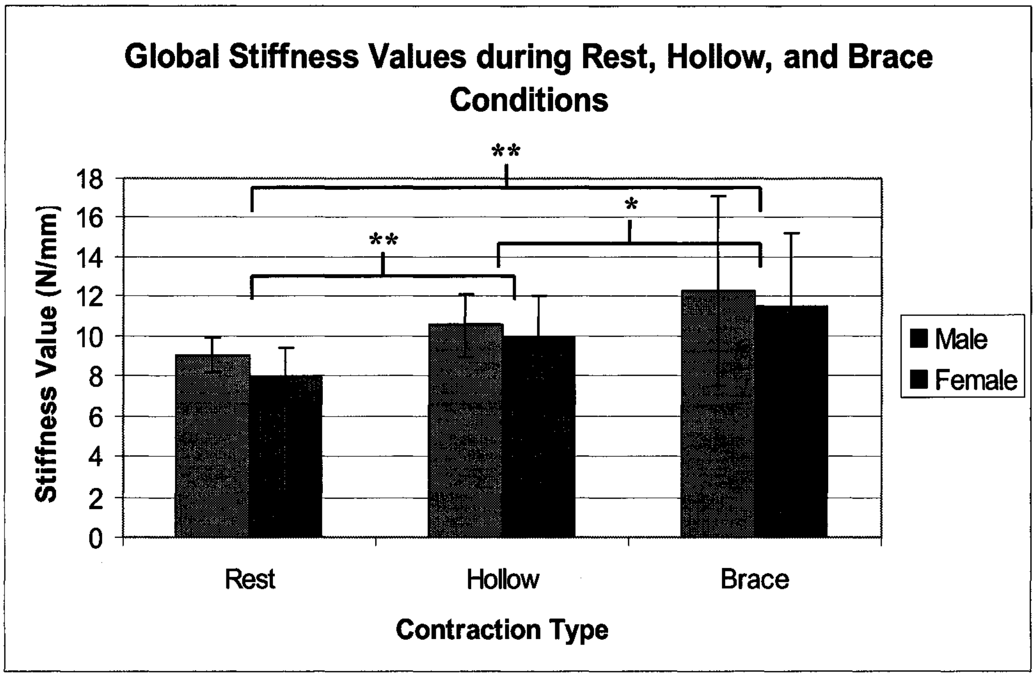

Figure 24: Mean (+/- SD) GS values for Rest, Hollow, and Brace Conditions; male and female results. No gender differences were found. * $p < 0.05$; ** $p < 0.01$.

Contraction Type	Percentage of Increase of Mean GS from Resting Mean GS Value	
	Male	**Female**
Hollow	16.6	24.6
Brace	36.2	43.9

Table 20: Percentage of increase of mean GS from the resting mean GS. Percentage of increase = (GS during hollow or brace – resting GS)/resting GS. GS measured in N/mm.

Mean maximal stiffness values exhibited similar results to those of GS values, demonstrating a significant effect for the experimental condition (rest, hollow, brace). (See Table 21) In particular, both the hollow and the brace contraction produced significantly higher MMS values than the MMS values taken during rest ($p < 0.00$). Again, the brace contraction produced significantly higher MMS values than the hollow contraction ($p = 0.01$). (See Table 22, Figure 25). No gender differences occurred

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between the MMS values during any of the three conditions (p = 0.78) and no significant effects of the covariates of age and BMI were seen (p = 0.852 and 0.732, respectively). Finally, a 25.9% increase in MMS values (from resting MMS value) occurred during the hollow contraction and a 36.5% increase occurred during the brace contraction. (See Table 23 for gender-specific percentages).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Condition	132.16	1.54	86.06	23.96	0.00*
Condition X Gender	1.58	1.54	1.03	0.29	0.69
Error (exercise)	143.39	39.93	3.59		

Table 21: Repeated Measures ANOVA for MMS values for rest, hollow, and brace conditions; male and female. MMS measured in N/mm. Sphericity not assumed; Huynh-Feldt adjustment used.

(I) Condition	(J) Condition	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	**Upper Bound**
Rest	Hollow	-1.82*	0.28	0.00	-2.40	-1.24
	Brace	-3.05*	0.54	0.00	-4.17	-1.94
Hollow	Rest	1.82*	0.28	0.00	1.24	2.40
	Brace	-1.24*	0.47	0.01	-2.19	-0.28
Brace	Rest	3.05*	0.54	0.00	1.94	4.17
	Hollow	1.24*	0.47	0.01	0.28	2.19

Table 22: Pair-wise comparison between MMS Values for the rest, hollow, and brace conditions. MMS measured in N/mm. Based on estimated marginal means. * Mean difference is significant at the p < 0.05 level.

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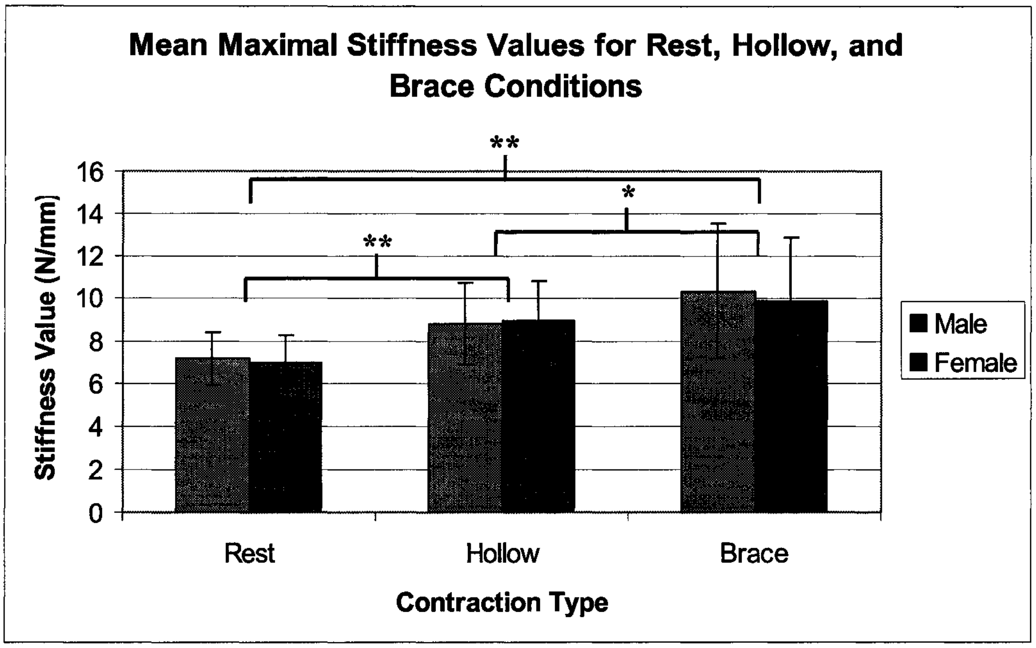

Figure 25: Mean (+/- SD) MMS values for Rest, Hollow, and Brace Conditions, male and female results. No gender differences were found. * $p < 0.05$; ** $p < 0.01$

Contraction Type	Percentage of Increase of the Mean MMS from Resting Mean Value	
	Male	Female
Hollow	22.4	29.3
Brace	30.7	42.3

Table 23: Percentage of increase of mean MMS from the resting mean MMS. Percentage of increase = (MMS stiffness during hollow or brace – resting MMS)/resting MMS. MMS measured in N/mm.

4.C.4 Analysis of the average normalized EMG values for trunk muscles for rest, hollow, and brace conditions

Repeated measures ANOVA demonstrated significant differences in muscle activity of the abdominal musculature (RA, EO, and IO) and the erector spinae musculature (T-ES L-ES) during the three conditions. (See Table 24, 25, 26, 27, and 28, respectively) For the abdominal musculature, pair-wise comparisons demonstrated significant differences between all three conditions ($p = 0.01$ for RA, $p < 0.00$ for EO, IO). (See Table 29 for

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RA, Table 30 for EO, and Table 31 for IO) Conversely, the EMG values of the erector spinae musculature (T-ES and L-ES) at rest were not significantly different than EMG values during the hollow condition. However, the rest and brace conditions demonstrated significant differences as did the hollow and brace conditions. (See Table 32 and 33). Please refer to Figure 26 for gender-specific EMG values for the rest, hollow, and brace conditions for both abdominal and erector spinae musculature. No gender differences were found between EMG values for the three conditions for any of the trunk muscles (p = 0.66, 0.67, 0.73, 0.19, 0.94 for RA, EO, IO, T-ES, and L-ES, respectively).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
RA	81.33	1.20	67.87	14.99	0.00*
RA X Gender	5.10	1.20	4.26	0.94	0.36
Error (RA)	141.08	31.16	4.53		

Table 24: Repeated Measures ANOVA for RA EMG values for rest, hollow, and brace conditions; male and female. EMG activity (Volts) expressed as a % of MVC. Sphericity not assumed; Huynh-Feldt adjustment used.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
EO	5802.93	1.15	5038.89	57.35	0.00*
EO X Gender	1.22	1.15	1.06	0.01	0.94
Error (EO)	2630.68	29.94	87.86		

Table 25: Repeated Measures ANOVA for EO EMG values for rest, hollow, and brace conditions; male and female. EMG activity (Volts) expressed as a % of MVC. Sphericity not assumed; Huynh-Feldt adjustment used.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
IO	16978.00	2	8489.00	46.30	0.00*
IO X Gender	136.95	2	68.48	0.37	0.69
Error (IO)	9533.85	52	183.34		

Table 26: Repeated Measures ANOVA for IO EMG values for rest, hollow, and brace conditions; male and female. EMG activity (Volts) expressed as a % of MVC. Sphericity assumed.

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Source	Type III Sum of Squares	df	Mean Square	F	Sig.
T-ES	1882.68	1.05	1786.32	25.00	0.00*
T-ES X Gender	102.57	1.05	97.32	1.36	0.26
Error (T-ES)	1958.08	27.40	71.46		

Table 27: Repeated Measures ANOVA for T-ES EMG values for rest, hollow, and brace conditions; male and female. EMG activity (Volts) expressed as a % of MVC. Sphericity not assumed; Huynh-Feldt adjustment used.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
L-ES	7157.10	1.05	6822.80	39.82	0.00*
L-ES X Gender	1.14	1.05	1.09	0.01	0.944
Error (L-ES)	4672.68	27.27	171.32		

Table 28: Repeated Measures ANOVA for L-ES EMG values for rest, hollow, and brace conditions; male and female. EMG activity (Volts) expressed as a % of MVC. Sphericity not assumed; Huynh-Feldt adjustment used.

(I) RA	(J) RA	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-0.46*	0.16	0.01	-0.79	-0.14
	Brace	-2.28*	0.53	0.00	-3.37	-1.19
Hollow	Rest	0.46*	0.16	0.01	0.14	0.79
	Brace	-1.82*	0.52	0.00	-2.90	-0.74
Brace	Rest	2.28*	0.53	0.00	1.19	3.37
	Hollow	1.82*	0.52	0.00	0.74	2.90

Table 29: Pair-wise comparisons between RA EMG Values for the rest, hollow, and brace conditions. EMG activity (Volts) expressed as a % of MVC. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

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(I) EO	(J) EO	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-2.24*	0.58	0.00	-3.42	-1.06
	Brace	-18.65*	2.29	0.00	-23.34	-13.95
Hollow	Rest	2.24*	0.58	0.00	1.06	3.42
	Brace	-16.40*	2.30	0.00	-21.13	-11.68
Brace	Rest	18.65*	2.29	0.00	13.95	23.34
	Hollow	16.40*	2.30	0.00	11.68	21.13

Table 30: Pair-wise comparisons between EO EMG Values for the rest, hollow, and brace conditions. EMG activity (Volts) expressed as a % of MVC. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

(I) IO	(J) IO	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-13.21*	2.71	0.00	-18.78	-7.63
	Brace	-34.51*	4.12	0.00	-42.99	-26.03
Hollow	Rest	13.21*	2.71	0.00	7.63	18.78
	Brace	-21.30*	3.86	0.00	-29.24	-13.36
Brace	Rest	34.51*	4.12	0.00	26.03	42.99
	Hollow	21.30*	23.86	0.00	13.36	29.24

Table 31: Pair-wise comparisons between IO EMG Values for the rest, hollow, and brace conditions. EMG activity (Volts) expressed as a % of MVC. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

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(I) T-ES	(J) T-ES	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-0.15	0.20	0.47	-0.57	0.27
	Brace	-10.12*	2.05	0.00	-14.33	-5.90
Hollow	Rest	0.15	0.20	0.47	-0.27	0.57
	Brace	-9.97*	1.95	0.00	-13.98	-5.95
Brace	Rest	10.12*	2.05	0.00	5.90	14.33
	Hollow	9.97*	1.95	0.00	5.95	13.98

Table 32: Pair-wise comparisons between T-ES EMG Values for the rest, hollow, and brace conditions. EMG activity (Volts) expressed as a % of MVC. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

(I) L-ES	(J) L-ES	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	0.44	0.26	0.11	-0.10	0.98
	Brace	-19.36*	3.17	0.00	-25.87	-12.85
Hollow	Rest	0.44	0.26	0.11	-0.98	0.10
	Brace	-19.80*	3.03	0.00	-26.02	-13.58
Brace	Rest	19.36*	3.17	0.00	12.85	25.87
	Hollow	19.80*	3.03	0.00	13.58	26.02

Table 33: Pair-wise comparisons between L-ES EMG Values for the rest, hollow, and brace conditions. EMG activity (Volts) expressed as a % of MVC. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

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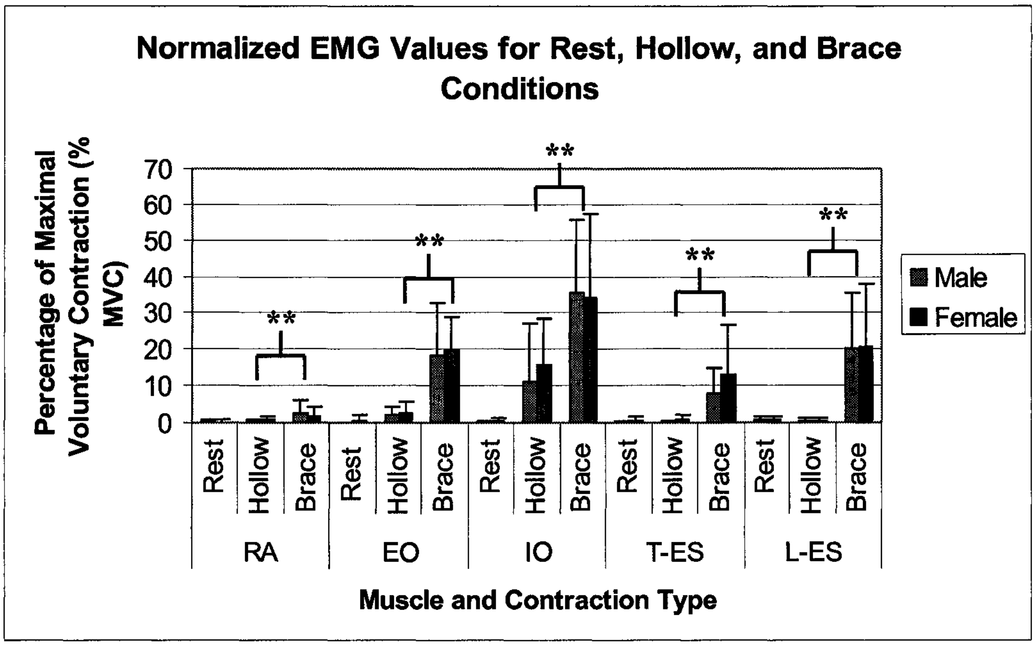

Figure 26: EMG values for trunk muscles for the rest, hollow, and brace conditions; male and female. Significant differences were present between the rest and hollow condition for RA, EO, and IO. Significant differences were present between the rest and the brace condition for all muscles. For clarity, only significant differences between the hollow and brace condition are shown. No gender differences were found. ** $p < 0.01$

4.C.5. Analysis of the average transversus abdominis cross-sectional area and mean thickness for rest, hollow, and brace conditions

The results of the repeated measures ANOVA demonstrated a significant difference in TrA CSA values during the three conditions (rest, hollow, and brace). (See Table 34) Pair-wise comparison exhibited significant differences for the TrA CSA when comparing the rest condition to the hollow and the rest to the brace ($p < 0.00$). However, no significant difference in TrA CSA was found between the hollow and the brace ($p = 0.10$). (See Table 35 and Figure 27) Finally, no gender differences in TrA CSA for any of the conditions were demonstrated ($p = 0.20$).

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Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Condition	72.84	2	36.42	53.86	0.00*
Condition X Gender	2.37	2	1.19	1.75	0.18
Error (rest)	35.17	52	0.68		

Table 34: Repeated Measures ANOVA for TrA CSA values for rest, hollow, and brace conditions; male and female. TrA CSA measured in cm^2. Sphericity assumed. * Significant at the $p < 0.05$ level.

(I) Condition	(J) Condition	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-1.73*	0.16	0.00	-2.05	-1.41
	Brace	-2.15*	0.24	0.00	-2.65	-1.66
Hollow	Rest	1.73*	0.16	0.00	1.41	2.05
	Brace	-0.42	0.25	0.10	-0.94	0.09
Brace	Rest	2.15*	0.24	0.00	1.66	2.65
	Hollow	0.42	0.25	0.10	-0.09	0.94

Table 35: Pair-wise comparisons for TrA CSA values for the rest, hollow, and brace conditions. TrA CSA measured in cm^2. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

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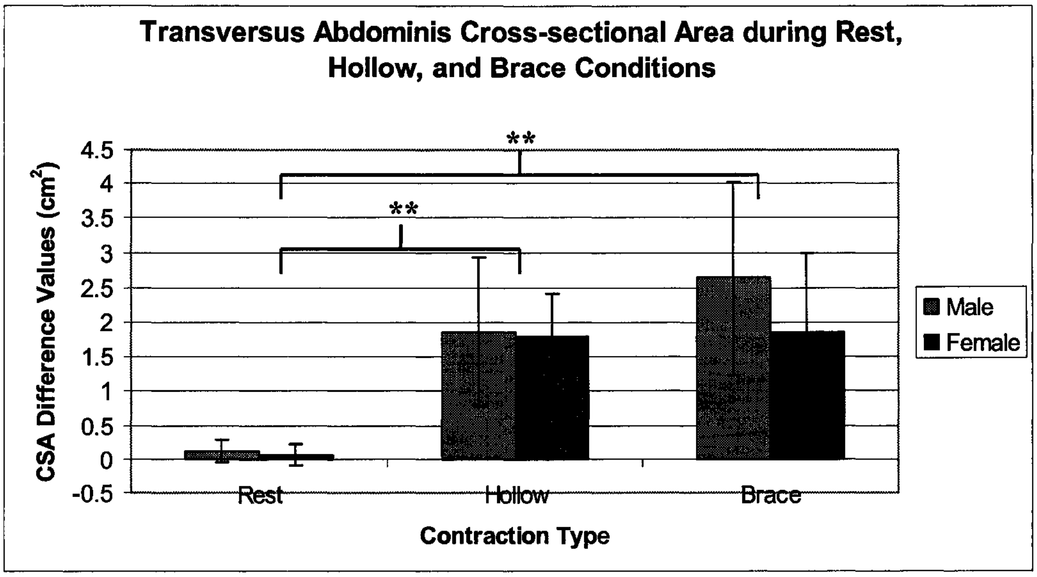

Figure 27: Cross-sectional area of transversus abdominis during the rest, hollow, and brace conditions, male and female results. Difference values represent the difference between the two ultrasound images taken before and during indentation during each condition. CSA measured in cm^2. No significant differences noted between genders. ** $p < 0.01$.

Similar results were found for the TrA mean thickness; repeated measures ANOVA again found a significant difference present in TrA mean thickness values during the three conditions. (See Table 36) Pair-wise comparison demonstrated significant differences between the rest condition and the hollow condition ($p < 0.00$) and the rest to the brace condition ($p < 0.00$). No significant difference was present between the hollow and the brace ($p = 0.15$). (See Table 37 and Figure 28). Further, gender was not found to be significant for TrA mean thickness ($p = 0.14$).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Condition	4.12	2	2.06	64.30	0.00*
Condition X Gender	0.13	2	0.07	2.10	0.13
Error (rest)	1.66	52	0.03		

Table 36: Repeated Measures ANOVA for TrA mean thickness values for rest, hollow, and brace conditions; male and female. TrA mean thickness measured in cm. Sphericity assumed. * Significant at the $p < 0.05$ level.

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(I) Condition	(J) Condition	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	Upper Bound
Rest	Hollow	-0.42*	0.04	0.00	-0.50	-0.34
	Brace	-0.51*	0.05	0.00	-0.60	-0.41
Hollow	Rest	0.42*	0.04	0.00	0.34	0.50
	Brace	-0.09	0.06	0.15	-0.21	0.03
Brace	Rest	0.51*	0.05	0.00	0.41	0.60
	Hollow	0.09	0.06	0.15	-0.03	0.21

Table 37: Pair-wise comparisons for TrA mean thickness values for the rest, hollow, and brace conditions. TrA mean thickness measured in cm. Based on estimated marginal means. * Mean difference is significant at the $p < 0.05$ level.

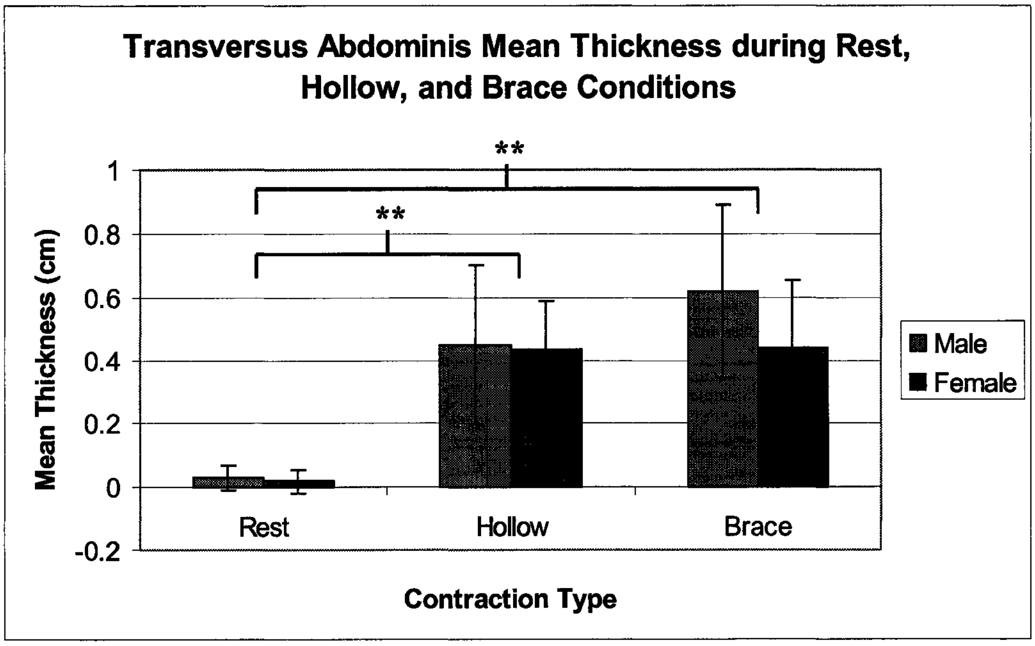

Figure 28: Mean thickness of transversus abdominis during the rest, hollow, and brace conditions; male and female results. Difference values represent the difference between the two ultrasound images taken before and during indentation during each condition. Mean thickness measured in cm. ** $p < 0.01$.

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4.C.8 Testing of confounders: Change in stiffness values over time and the effect of order of contraction

Results of paired t-tests did not demonstrate a significant difference between stiffness values measured during the 2nd and 10th indentation ($p = 0.32$, 0.43 for GS, and $p = 0.13$, 0.48 for MMS, for males and female respectively). This suggests that changes in stiffness values over time did not occur for either stiffness measure. Further, there was no effect of the order of contraction (performing hollow contraction first versus brace contraction first). Using a repeated measures ANOVA, the order of exercise performance did not demonstrate significance ($p = 0.73$ for GS and $p = 0.68$ for MMS).

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CHAPTER FIVE
DISCUSSION

5.A Experiment One

The first hypothesis of this research predicted that AI would demonstrate excellent reliability (ICC $\geq$ 0.75).[130] The results of this study support this hypothesis as AI exhibited excellent intra-rater reliability for all outcome variables used to quantify L4 stiffness (ICC values of 0.91 for GS and 0.93 for MMS). Further, inter-trial inconsistency remained below 10% for all stiffness variables.

The ICC values found for the assessment of spinal stiffness obtained from the AI technique were much higher than those found for manual testing of spinal stiffness (using therapist's hands). Overall, reliability values for the evaluation of spinal stiffness using the PA pressure test have been found to be poor.[88-90] Matyas and Bach [90] first found poor reliability of manual PA stiffness assessment, reporting Pearson's r ranging from 0.09 to 0.46. Later studies also noted poor reliability, [88, 89] with ICC (1,1) values ranging from 0.03-0.37. With improvements to the testing protocol and delineation of stiffness into ranges, reliability increased to a fair level [130] with an ICC value reported to be 0.55 (range 0.50-0.62).[92] The ICC value of the PA pressure test increased further when an 11-point stiffness rating scale was employed and more rigorously controlled testing protocol were used (ICC = 0.77).[92] Although improvements in the reliability of the manual assessment of spinal stiffness have been demonstrated, it is only under standardized, but artificial, conditions that are not typically employed in the clinical environment.

The observation that AI exhibits greater reliability than manual assessment of spinal stiffness is expected for two main reasons. First, AI measures several variables in an objective manner, increasing the reliability of spinal stiffness assessment. Specifically, use of technology to quantify force and displacement data (load cell and a LVDT, respectively), in addition to customized computer programming, allows consistency of force application and real-time visualization of results. These developments allow

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objective quantification of spinal stiffness while manual techniques rely on cognitive recollection of stiffness levels for a subjective impression of stiffness.

Another factor that explains the increased reliability of AI is the limitations present in manual stiffness assessment. It has been well documented that certain factors, such as visual occlusion, [141] peak force, [43] frequency of PA loading, [97, 142] and direction of force application, [143] can affect estimates of stiffness measured manually. When these factors are not controlled at the time of manual assessment of spinal stiffness, they work to decrease the reliability of stiffness measurements. With AI, these factors are effectively controlled through use of real-time measurement of force and indentation frequency data such that continuous feedback to the operator is present. Further, with AI the angle of indentation is kept constant (which is more easily assessed using an instrument than when direction of force application is subjectively judged in manual assessment). Most importantly, it has been demonstrated that the forces used by therapists during PA mobilization are extremely variable among clinicians applying the same manual technique.[144] This finding alone could explain the discrepancies in stiffness values obtained when performing manual testing as compared to AI. As mentioned above, the real-time assessment of applied force, in conjunction with the ability to set a maximum force limit, allows for consistency in force application to be accomplished with AI.

On the other hand, the reliability values for AI, although slightly lower, are comparable to those found for automated indentation devices. Intra-class correlation coefficient values have been reported to be over 0.90 for almost all automated indentation instruments. Specifically, the SPAM was found to have an ICC value of 0.979 at L5,[99] Lee and Evans' stiffness assessment device had an ICC value of 0.99 for L3/4 and 0.95 for L4/5,[98] SAM had an ICC value of 0.96 for lumbar vertebrae,[44, 45, 93] and Rigid Frame Indentation at 0.99-1.00 for varying experimental conditions.[100] Interestingly, the reliability of AI was higher than that of the SPS which found an ICC value of 0.88 at L3.[97] That automated indentation devices have higher reliability values (overall) than AI is not surprising. It is known that the stiffness of a viscoelastic material is dependent upon the velocity at which force is applied.[37] Therefore, although the rate of indentation in this study (using AI) was

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standardized using a visual cue, slight variations in the rate of indentation were likely to occur. While these variations were not of sufficient magnitude to create a situation where reliability was poor, they may account for the slightly lower reliability values that occur with AI compared to other automated techniques. In the case of automated indentation, no variation in indentation rate is present.

The suspicion that variance in the indentation rate affects stiffness measures is further supported by the findings of the paired t-test. A significant difference was present between indentation 2 and 10 for GS values, but not for MMS. This suggests that in GS, stiffness values do not stay consistent over time and visual graphical analysis demonstrates a trend for the GS values to increase in stiffness over time (See Figure 21, Chapter 4). Because GS is calculated by fitting a regression line to the force-displacement curve from 30N to maximal force, any variations in indentation rate would affect the GS values to a greater extent than the MMS values. As the MMS values only take into account the maximal force imparted and the maximal displacement, they are less dependent on overall rate of indentation. While high correlation coefficients were demonstrated for both GS and MMS values, in the case of GS, high correlation coefficients suggest a proportional relationship between indentation 2 and 10. Further, due to the relatively small absolute difference between the 2nd and 10th GS value (approximately 0.40 N/mm), it is not known if GS variation related to successive indentation is clinically significant. These results should be placed in context with the contradictory results in Experiment Three.

It should be noted that large differences in individual subject inter-trial inconsistency values were exhibited, with some subjects having ITI values approaching 30% (+/- 1 SD). This suggests that the consistency of stiffness results obtained by AI may be specific to the individual and may be influenced by other factors not defined in this study. One possible confounding factor that could explain the measurement inconsistency with certain subjects may be movement of the indentation contact point on the subject's spine or failure to control subject specific factors which influence stiffness (eg. IAP, muscle contraction, etc.).[35] Even small changes in the subjects' positioning could result in force

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being applied in an altered direction. In this situation, changes in measured spinal stiffness may occur as the indentation test may involve different anatomy. In addition, previous research by Allison et al.[145] demonstrated a significant difference in the stiffness of the spine when the angle of indentation was altered. Accordingly, should the angle of indentation change between trials, variability of the measured outcomes may increase. Finally, the subject's baseline stiffness could also be a confounding factor. Although a formal analysis was not performed, it was observed that those subjects with high baseline stiffness values for GS and MMS (stiff back) often had large changes in their stiffness values over time.

5.B Experiment Two

The second hypothesis of this study stated that the cross-sectional area measurement of TrA during contraction would demonstrate excellent reliability (ICC $\geq$ 0.75). Results from the present study support this hypothesis with ICC values for measurement of TrA CSA calculated to be 0.998. This reliability is excellent [130] and is comparable to reliability values from a previous study by Teyhen et al., [123] where the ICC for intra-image, intra-rater reliability was 0.98 and inter-image reliability had an ICC value of 0.93. Kidd et al.[125] also demonstrated similar reliability for TrA thickness measures when testing different patient positions (sit, stand). Intra-class correlation coefficient values ranged from 0.90 – 0.96 (SEM 0.29 – 0.57mm). When comparing between days, the ICC values remained high, with standing positions having an ICC of above 0.96 and sitting of 0.88 (SEM 0.18-0.33mm).[125] In addition, Stokes et al.[128] demonstrated ICC results for intra-rater reliability of the CSA measurement of the multifidus muscle ranging from 0.98 and 1.00.

It should be noted that the studies of Teyhen et al.[123] and Kidd et al.[125] utilized TrA thickness as the main outcome measure. However, because CSA was used to measure TrA contraction in the present study, direct comparison to the studies of Teyhen et al.[123] and Kidd et al.[125] should be made with caution. Similarly, the study by Stokes et al.[128] used the CSA measurement technique, but in a muscle different to the one imaged in this study. Further, all the above studies examined the measurement of TrA anteriorly on the

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trunk, while the present study measured TrA posteriorly. Therefore, there is no single study which is directly comparable to the one in this project. However, by general comparisons, it would appear that the reliability in this study is reasonable, given the results obtained by other investigators in similar, but not equal circumstances.

Due to the differences outlined above between the present study and the previous literature, a secondary speculation was made regarding the validity of the measurement of TrA contraction in this study. Specifically, it was theorized that the TrA activation ratio (CSA of TrA contracted divided by the CSA of TrA at rest) would be somewhat smaller, although comparable, to previously calculated TrA ratios of trained subjects performing the hollow contraction (where imaging occurred in the anterolateral approach and a thickness measurement of TrA was used). This speculation was generated as the subjects of this present study received less training in the hollow contraction than subjects in previous studies. Specifically, the subjects in the present study underwent TrA contraction training and testing on the same day whereas other studies often utilized a two day testing design where training occurred the first day and formal testing the second.[59] This may result in larger TrA contractions (causing a larger TrA activation ratio) due to decreased levels of TrA fatigue in subjects that were trained over two days. The present study chose to use a one day testing design to minimize subject attrition.

Further, studies have also used increased overall training time as compared to the training time used in the present study (for example, five TrA contractions in three different positions, followed up by specific ultrasound feedback training for approximately 5 minutes in three different positions).[123] In previous studies using this type of training, the activation ratio was ≥ 2.0 in trained subjects.[123] Increased training time may improve the quality of TrA contraction (directly due to more practice and feedback) resulting in larger TrA activation ratios. However, while fatigue did not appear to occur in the study by Teyhen et al., high numbers of repetitions of the TrA have been shown to cause fatigue in subjects not previously trained in the TrA contraction.[34] In the present study, increasing training time was not considered to be a viable option as subjects were required to not

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only hold, but maintain, TrA contraction against force (indentation). Therefore, the minimization of TrA fatigue was considered important.

The validity speculation was met in the present study as the TrA activation ratio was not found to be significantly different than the previously established activation ratio ($p = 0.21$ and 0.39 for males and females, respectively). Further, as theorized, the activation ratio was found to be slightly less than the established ratio, measured to be 1.83 in males and 1.89 in females.

Lastly, through simulation of error situations possible in the measurement of thickness and CSA on ultrasound images, it was seen that with the type of images present in this study (TrA angulated and curved), CSA measurement exhibited decreased levels of percent error than did thickness measurements. This, in combination with excellent reliability and validity findings, supports the use of the CSA measurement technique to quantify TrA contraction.

5.C Experiment Three

5.C.1 Comparison of stiffness values of the rest condition to the hollow and brace conditions

The third hypothesis of this research stated that both the hollow and the brace contractions would significantly increase the stiffness of the spine as compared to stiffness at rest. This prediction was based on previous research which established that contraction of the trunk muscles (abdominals and erector spinae) increase stiffness of the spine[15, 95, 146, 147] and SI joints.[59]

This hypothesis was supported by the results of this thesis as it was demonstrated that the stiffness values obtained during the hollow and the brace contractions were significantly larger than the stiffness values obtained at rest ($p \leq 0.000$). This result remained consistent regardless of gender or the order in which the contractions were performed. These results are consistent with previous studies that demonstrated increases in stiffness of the spine with the addition of muscular contraction. In a study by Lee et al.,[147] the

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stiffness of the spine at L3 was examined during a maximal voluntary back extensor contraction. The mean increase in PA stiffness during MVC was found to be 350%, confirming that muscular contraction does work to substantially increase stiffness when performed maximally. Similar results were found by Colloca and Keller[95] where stiffness was measured using high loading rate PA manipulative thrusts at the L3 spinous process during isotonic lumbar extension. An increase ranging from 1.3% - 39.4% occurred during the trunk extension tasks as compared to the apparent mass at rest.[95] This supports the current study's finding that stiffness increases during muscular contraction of the trunk.

It should be noted that the above studies focused on maximal trunk extension efforts, while the present study evaluated trunk contraction levels at a much lower level (15-30% MVC). Therefore, comparison of this experiment's results to studies evaluating the role of smaller amounts of muscle activity on spinal stiffness is warranted. Specifically, Shirley et al.[146] used the Spinal Physiotherapy Simulator to quantify the stiffness of L4 during different levels of back extensor muscle contraction. An 11.8% increase in mean stiffness (as compared to resting stiffness value) was demonstrated when a 10% MVC contraction was used and a 41.2% increase in stiffness occurred when 30% of MVC contraction was performed.[146] These results compare favorably to the increase in stiffness levels found in the present study where a 25.9% increase in MMS occurred during the hollow contraction and 36.5% during the brace contraction (genders combined). For GS, levels of stiffness increased by 20.6% for the hollow and 40.1% for the brace. As expected, these percent increases in stiffness values fall between the percent increase values documented by Shirley et al., however, the absolute values of stiffness in the present study were lower. Compared to Shirley et al., [146] who found that mean stiffness levels increased from 14.8 N/mm to 17.5 N/mm and 21.9 N/mm with a 10% and 30% of MVC for the ES (respectively), results from our study found baseline stiffness levels at 8.50 N/mm and 7.05 N/mm, 10.23 N/mm and 8.86 N/mm for the hollow, and 11.88 N/mm and 10.10 N/mm for the brace (GS and MMS values, respectively).

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There are several reasons as to why the absolute values of stiffness from this project were lower than those found by Shirley et al. First, differences in the method of assessing stiffness between the two studies are evident (AI versus SPS). Specifically, the testing frequency and indentor head size were different between studies making results difficult to compare as these factors are known to affect spinal stiffness.[148] Second, differences in the type of muscle contraction performed particularly in muscle activation (abdominal and back muscle contraction versus back extensor contraction only), likely account for some of the incongruity found in both percentage increases and absolute stiffness levels. It has been reported that during lumbar extension, a greater approximation of the articular surfaces of the zygapophyseal joint occur, increasing the resistance to anterior displacement.[149] Shirley et al.[146] recognized that although the lumbar extension in their study was intended to be isometric, it was possible that a physiological extension of the lumbar spine could occur. This may partially explain the larger increases from baseline stiffness that occurred during lumbar extension in their study as compared to the present study. Overall, because numerous factors exist that may contribute to differences in results between the present study and Shirley et al.'s work, it is significant that the stiffness values between the studies were similar.

Although examining the impact of different levels of muscular contraction on spinal stiffness, the above mentioned studies do not investigate the particular contractions used in the present study. Consequently, a more specific comparison to the hollow and brace contractions used in the present study is required. Richardson et al., [59] albeit testing the sacroiliac joint, utilized both the hollow and brace muscle contraction. Similar to the results of the present study, Richardson et al. found that both the hollow and the brace contractions decreased the laxity value (increased the stiffness) of the SI joint.[59] Again, specific comparison of stiffness values between the studies may not be valid as the techniques to quantify stiffness were significantly different (vibration analysis using Color Doppler ultrasound versus AI) as was the location of stiffness measurement (SI joint versus lumbar spine). However, it still remains that the same muscular contractions induced increases in stiffness values (decreased laxity values) in the SI joint, as they also did in the spine in the present study. Because similar muscles that control spinal

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movement also insert anatomically around the SI joint, [51] it is reasonable to expect that similar findings regarding an increase in stiffness with muscle contraction would occur both in the SI joint and the lumbar spine.

While comparison to the Richardson study is helpful, a study by Hodges et al.[15] allows a more specific comparison to the work performed in this experiment. Using evoked contractions of TrA in an *in-vivo* porcine model, the relative intervertebral displacement of both the L3 and L4 vertebrae was reduced and the stiffness of L4 in a sagittal plane was increased for caudal displacements by 16%.[15] This stimulation of the TrA muscle closely relates to the hollow contraction used in the present study whereby selective recruitment of TrA and multifidus was performed. In the Hodges et al. study, [15] stiffness was quantified as the slope of the regression line fitted to the force-displacement data of L4, similar to our GS outcome measure used in this project. This 16% increase in stiffness found by Hodges et al. compares favorably to the GS value obtained in the present study where males increased their stiffness of L4 by 16.6% and females by 24.6%. Therefore, in both cases, increased stiffness levels (as compared to rest stiffness) occur with a contraction of the TrA musculature (hollow contraction).

The higher levels of spinal stiffness found in the present study (particularly in females) as compared to Hodges et al. may have occurred due to numerous factors. Differences were present between the two studies in regards to the methodology of quantifying spinal stiffness; however, these differences would actually suggest that increased stiffness should have been found in Hodges et al.'s method. Specifically, Hodges et al. measured spinal stiffness using the force-displacement relationship of the L4 vertebra in the sagittal plane[15] whereas the present study quantified stiffness of L4 within the transverse plane. Due to the relatively small movements of vertebrae in the cephalad and caudad directions, it would be anticipated that higher stiffness values would occur with sagittal plane measurement of L4 stiffness. However, further investigation in the methodology of the two studies likely explains the differences. Hodges et al.[15] removed both the supraspinous ligament and the thoracodorsal fascia from the porcine specimens prior to stiffness testing, consequently removing anatomical restraints to cephalad-caudad

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movement in the vertebrae, resulting in lower levels of measured stiffness. Further, much smaller force levels were used by Hodges et al. than in the present study (1 – 7N versus 100N).[15] It has been established that stiffness of a tissue increases with increasing loads,[43] providing a strong rationale for the present study's higher stiffness values as compared to Hodges et al. Lastly, subjects performing the hollow contraction within the present study were also instructed in the contraction of the multifidus, whereas Hodges et al. stimulated only the TrA. This additional muscular contractive force may also have resulted in increased stiffness in L4 in the present study.

5.C.2 Comparison of the stiffness values of the hollow condition and the brace condition

The fourth hypothesis of the present research stated that the hollow contraction would increase stiffness of the spine to a greater extent than the brace contraction. This hypothesis was based on numerous areas of research. First, anatomical knowledge suggests that the muscles utilized in the brace contraction (general muscle contraction) are primarily "mover" muscles, such that a large contraction of these muscles facilitates trunk movement on the pelvis or vice versa.[21] Further, due to the location and alignment of the deeper TrA muscle and multifidus muscle, it was anticipated that a contraction using these muscles would be more effective at increasing spinal stiffness at a segmental level than the more general contraction found in the brace condition.[21, 54, 68] Given these factors and the muscles' demonstrated ability to increase intra-abdominal pressure (a known factor that increases stiffness of the spine), it was anticipated that TrA would increase stiffness of the spine to a greater extent.[52] Additionally, this hypothesis was based on studies performed on the SI joint where the laxity value was decreased to a greater extent by the hollow contraction than the general brace contraction.[59]

However, the results obtained in this study did not confirm the hypothesis that stiffness of the spine would be increased more by the hollow than the brace; in fact, the results of this study suggest that the brace contraction was more effective in increasing spinal stiffness. Stiffness measures (both GS and MMS) demonstrated an increased level of stiffness at the L4 vertebrae during the brace contraction as compared to the hollow contraction. This

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difference was statistically significant for both GS and MMS values ($p = 0.02$, and $p = 0.01$, respectively). While this experiment found bracing to be more stiffening than the hollow contraction, the following paragraphs will describe that this finding is valid.

Three studies have been identified that specifically compare the hollow and the brace contractions. These include studies performed by Richardson et al., [59] Grenier and McGill, [25] and Vera-Garcia et al.[78] The first, by Richardson et al.[59] demonstrated that the hollow contraction significantly decreased the SI joint laxity (increased stiffness) to a greater extent than with the brace contraction. In the case of Richardson et al., the incongruity of their results to those of the present study may be explained by the anatomical differences in the TrA muscle between the anatomic locations from which stiffness data were obtained. As mentioned previously, it is known from cadaveric studies that there are 3 muscle bands of the TrA, the upper, middle, and lower fascicles. It is thought that the lower fascicles of the TrA work primarily to stabilize the SI joint, while the middle fascicles work to stabilize the lower lumbar spine.[31] In this way, it may be possible that different parts of the TrA musculature are better at stabilizing different locations in the body whether due to biomechanics or increased efficiency in the contraction of certain muscle bands.

Further, the superficial, global muscles such as RA, EO, IO, and the erector spinae muscles most likely have more influence within the lumbar spine. The lumbar vertebrae are more mobile than the SI joint which is documented to have very little movement associated with it.[150] This means that a contracting muscle could potentially have more effect on the stiffness of the lumbar vertebrae than perhaps the SI joint which already has little movement available (ceiling effect). Also, anatomically, the superficial muscles cross over and around the spinal vertebrae more so than the SI joint, allowing the muscles in this area to have a greater biomechanical influence on the stiffness of the spinal vertebrae compared to the SI joint.[24]

While our findings are opposite to those of Richardson et al., they are in direct agreement with the findings of Grenier and McGill.[25] In their study, a computational model was

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created that took into consideration the forces exerted by the abdominal and back muscles on the spine. Using EMG data from the respective muscles, a stability index was calculated that allowed spinal stability during certain tasks to be expressed. In this case, a higher stability index was indicative of greater stability. Four conditions (no load, two-hand load, left hand load, and right hand load) during the two contractions (hollow and brace) were tested. It was found that the stability index values differed significantly between the hollowing and bracing conditions (bracing > hollowing, $p = 0.001$) and between loading conditions ($p = 0.009$).[25] For all loading conditions, the brace increased the mean stability index (Nm/rad) to a greater extent than the hollow.[25] Further, using simulations of the contractions of the hollow and the brace at 20% of MVC (inputted into the spinal model), the brace was calculated to improve stability over the hollow by 32%.[25] Support for this finding of the brace contraction increasing stability more than the hollow contraction occurs in research by Crisco and Panjabi, albeit dealing with stability in the frontal plane, where it was found that global muscles (as used in the brace contraction) provide better stabilization to the spine.[24] Similarly, in the present study, the brace contraction increased the stiffness of the spine more so than the hollow contraction, although the percentage of increase in stiffness due to the brace above that obtained by the hollow was lower, calculated to be 19.5% for GS values, and 10.7% for MMS values. It has been documented that stiffness of the spine and muscles surrounding the spine creates a stabilizing effect.[151] In this way, the findings of our study relate very closely to the findings of Grenier and McGill.

Our findings of a smaller percentage of increase in stability as compared to the results of Grenier and McGill[25] are likely explained due to differences inherent in our method of assessing stability. In the present study, we calculated the specific force-displacement properties of one lumbar vertebrae with varying types of muscle contraction and in Grenier and McGill's study, the equilibrium state of the entire spine was assessed. Additionally, Grenier and McGill utilized simulation conditions whereby a specific percentage of MVC was assigned to muscles and inputted in the spinal model to calculate a resultant spinal index. This results in completely controlled contraction performances that were not achievable in the present study using *in-vivo* human performances.

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The last study that specifically compared the effects of the hollow and the brace contraction on spinal stability was performed by Vera-Garcia et al.[78] In their study, the effect of the hollow and the brace contractions on the control of spine motion and spinal stability as a response to sudden trunk perturbations was compared. A lumbar spine model was once again utilized to quantify spinal stability. It was found by Vera-Garcia et al. that both the stability index and compression levels of the spine (N) were statistically higher ($p < 0.0001$) for the brace contraction than the hollow contraction for the three levels of muscle activation (10%, 15%, and 20% of MVC).[78] Further, the brace condition significantly reduced the amount of lumbar displacement during sudden perturbation when compared to the hollow contraction.[78] Again, the findings of Vera-Garcia et al. support the results of the present study by further establishing the superior effect of the brace contraction in enhancing spinal stability.

While one can compare this experiment's results to any of the three related papers discussed above, it must be noted that this is the first study to investigate stiffness in a single lumbar location comparing the hollow and brace contractions. Therefore, these are new results which have no parallel in the literature and as such, provide new insight into the mechanics of spinal function.

Verification of the muscle activity in the three experimental conditions

In order to increase confidence in the results of this study, it must be established that two separate muscular contractions did occur. This involves an analysis of both the EMG data for superficial muscle activity and the ultrasound data for TrA contraction. Please see Table 38 for a summary of the expected and observed contractions of the trunk muscles for each of the three experimental conditions.

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	Rest			Hollow			Brace		
	TrA (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)	*TrA* (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)	*TrA* (CSA)	*RA, EO, IO* (EMG)	*T-ES, L-ES* (EMG)
Expected	None	None	None	↑	Min.	Min.	↑	↑	↑
Observed	None	None	None	↑	Min.	None	↑	↑	↑

Table 38: Expected and observed contraction of the trunk muscles for each condition: rest, hollow, and brace. TrA – transversus abdominis; RA – Rectus Abdominis; EO – External Obliques; IO – Internal Obliques; T-ES – Thoracic Erector Spinae; L-ES – Lumbar Erector Spinae; Min. – minimum contraction.

In regards to the superficial abdominal muscles, EMG values were significantly larger during the brace condition than the hollow condition, and EMG values in the hollow condition were significantly larger than the rest condition, as was expected. These results suggest that three different experimental conditions did exist. Further, the EMG findings during the conditions of rest, hollow, and brace are comparable to those discussed previously in the literature.[25, 59] Richardson et al. exhibited similar results with EMG activity during the hollow contraction significantly larger than at rest ($p < 10^{-4}$) and the brace contraction EMG activity significantly larger than during the hollow contraction ($p < 10^{-4}$).[59]

For the erector spinae musculature, significant differences were present between the hollow and the brace condition with the brace exhibiting higher EMG values. However, for the ES musculature, the rest and the hollow contractions did not demonstrate a difference. Literature on the performance of the hollow and brace contractions supports the present study's finding of a difference between ES EMG activity during the hollow and the brace.[59] However, previous literature has demonstrated a significant difference in ES EMG activity between the rest and hollow condition.[59] This difference in findings may be due to many factors. Simply, it may be that the subjects in the current study were better at maintaining relaxation of the ES musculature during the hollow contraction. Additionally, differences in stiffness quantification methodology between studies may affect the ability of the ES musculature to stay relaxed (vibration analysis versus AI).

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To further determine if the contractions of the hollow and the brace were performed correctly, analysis of the TrA contraction on B-mode ultrasound was completed. Specifically, significant differences were present between TrA CSA/mean thickness at rest and TrA CSA/mean thickness during the contractions (both the hollow and brace contraction). Further, no difference was found between TrA muscle size during the hollow and the brace contraction. This was expected based on previous studies.[59] Using B-mode ultrasound to quantify TrA contraction during performance of the brace and hollow condition, it was demonstrated (as per ultrasonic images within the paper) that contraction of TrA also occurs with the brace contraction as it does with the hollow contraction.[59]

Based on the results of the trunk muscle EMG values and TrA contraction level results, it can be hypothesized that the brace contraction increased the stiffness of the spine to a greater extent than the hollow contraction merely due to a larger number of muscles involved in the contraction. In the present study, the brace contraction involved the activation of the TrA in addition to all the other trunk muscles. Previous literature has demonstrated that the more muscle tension and stiffness that is present, the greater the stiffness of the spine.[146] This gives support to the findings that the stiffness of the lumbar spine was greater with the brace contraction. However, this finding may also be a function of the use of asymptomatic subjects. It has been reported in the literature that following a low back injury, TrA exhibits changes in its contraction performance, specifically alterations in timing.[16] It has also been reported that learning these contractions (especially the hollow) are very difficult for patients with low back pain to perform due to their inability to selectively recruit TrA.[152, 153] In theory, this may carryover to the performance of the brace, where in normal subjects for whom TrA contraction is not impaired, the TrA does contract with the brace condition, as seen in the results of the present study. However, in subjects with low back pathology, contraction of TrA may not occur when performing the brace contraction. Unfortunately, no studies exist to support or refute this suggestion.

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Conversely, this study's findings of the brace contraction increasing stiffness to a greater extent than the hollow contraction may be explained by a comparison of the relative levels of muscular contraction. As mentioned previously, the subjects were instructed to perform both the hollow and the brace at approximately 10-30% of their total maximal voluntary contraction. Although the majority of the average EMG values for the trunk muscles stayed below 30% of MVC, when examining the individual EMG data of the subjects, it was apparent that some subjects reached very high levels of their MVC when performing the brace contraction. For example, the MVC for IO during the brace for one female subject was 73.53% of MVC and for one male was 50.00% of MVC for lumbar ES. Because this study looked to evaluate the clinical teaching of these two contractions, subjects that "over-contracted" during the brace condition were not excluded, as their contractions were representative of what would occur in a clinical environment. However, this potentially results in a difference in the muscle contraction levels between the hollow and the brace. While a direct percentage of MVC could not be calculated for TrA due to limitations in the protocol, it has been suggested that only a small percentage of MVC is attainable for the TrA without other abdominal muscles also beginning to contract.[65] Further, it was known based on this study's results that the activity of the other superficial trunk muscles was low during the hollow condition compared to the brace pattern. Because the brace is a very general muscular activation of all abdominals and back muscles, it can occur at many different % MVC levels. Consequently, if subjects were contracting their superficial trunk muscles to such a high degree during the brace contraction, increased levels of stiffness ***should*** occur as compared to the hollow contraction, as a direct result of a higher percentage of MVC being utilized.[146] Nevertheless, because this "over-activation" occurred only in certain subjects (approximately four out of 28 subjects), it is unlikely that this was a determining factor in the results of this study. Significant differences were found to exist between the brace and hollow condition, suggesting that these higher % MVC contractions of certain subjects likely represent the variability demonstrated in the stiffness results of this experiment.

It must be noted, however, that each individual exhibited variation in the performance of the hollow and the brace contractions, both in activation of musculature and percentage

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of MVC recruited. Specifically, when performing the hollow contraction, some subjects were able to keep a virtually silent EMG record indicating no superficial abdominal muscle/erector spinae involvement. However, others were unable to separate the contraction of TrA with that of the internal obliques, or in some cases, the external obliques. Similar findings occurred with the brace contraction, with differences apparent in the level of contraction of each muscle group. Some subjects used the internal and external obliques primarily, while others utilized the erector spinae musculature predominantly. Additionally, a difference in the activation of the TrA muscle occurred during performance of the brace. Some subjects did not exhibit a contraction of TrA at all during performance of the brace (and had good levels of contraction for the superficial trunk muscles) while others had a larger TrA contraction during the brace than the hollow. This asymmetry in contraction performance could potentially explain differences in stiffness values achieved by the subjects and although not within the scope of this thesis, warrants further investigation and analysis in terms of how training quality or duration may affect contraction patterns. It may be that increased training increases the homogeneity of contractions. Further, more research is needed in regards to the possibility that people may have default contraction strategies that are difficult to reform. However, this finding of asymptomatic individuals having different muscle activation patterns is not unique to this study.[23, 154] Therefore, it must be acknowledged that the performance of the contractions remains specific to the individual.

Regardless of differences in muscle activation patterns within each subject, it remains that the brace contraction significantly increased spinal stiffness to greater extent than the hollow contraction. Again, the variations in muscle activation contribute to the variability in stiffness values achieved during performance of the hollow and the brace contraction. It is possible that with increased training time and/or further feedback to the subjects, these variations in spinal stiffness levels achieved with the respective muscle contractions would decrease and even greater differences between the hollow and the brace condition would be apparent.

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Clinical significance

The propensity to "over-activate" the trunk musculature when performing the brace contraction may be important to note when prescribing this exercise clinically or when making a clinical decision regarding treatment. It has been demonstrated that the brace contraction facilitates muscular co-contraction to a greater extent than the hollow contraction.[78] Further, it has been demonstrated that increased muscular co-contraction significantly increases the compression loads acting on the lumbar spine.[78] These increases in compressive loads have been linked to low back pain and low back disorders.[155, 156] This knowledge in combination with the present study's findings of some subjects' tendency to use high levels of muscle activation with the brace contraction raises issues regarding the prescription of this exercise. Therefore, within a clinical environment, the relative cost of performing the brace contraction must be weighed with its benefits. However, because this "over-activation" of the trunk musculature occurred only in relatively few subjects, and overall, the brace provided a significantly greater degree of stiffening, prescription of the brace contraction may be more dependant on the amount of education and training given to patients (teaching them not to over-activate the trunk muscles) rather than the concern of increased spinal compression.

Conversely, a major argument against prescription of the "abdominal hollowing" stabilizing contraction is the suggestion that the ability of humans to isolate the TrA is extremely rare. It has been estimated that following 1-2% of maximum voluntary contraction (MVC), the internal oblique is recruited with TrA along with the rest of the abdominal wall.[65] However, recent literature has shown support of the selective activation of TrA, using real-time ultrasound to both train and confirm its proper activation with minimal activation of rectus abdominis or the obliques.[59, 123, 127] Findings of the present study were similar, in that the majority of subjects were able to initiate a selective contraction of TrA with minimal muscle contraction of the superficial abdominals. This suggests that this selective contraction is possible with proper instruction.

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5.C.3 Address of possible confounding factors

One main factor was identified that may influence the extent to which the results of this study are valid. This factor was the constancy of stiffness values over time, as measured during the rest condition. If the stiffness values measured during the rest condition were found to change over time, it becomes difficult to ascertain whether changes in stiffness with muscular contraction were due to that contraction, or merely were more a function of time. Results of the study demonstrated that both measures of spinal stiffness, GS and MMS, did not exhibit a significant difference between the 2^{nd} and 10^{th} indentation (both during rest condition), strengthening the assumption that changes in spinal stiffness were due to muscle contraction. This observation is in conflict with the reliability results of Experiment One where differences in GS values were found to occur over time.

This difference in results may be explained by changes in viscoelastic properties of subjects within Experiment Three. The primary difference between Experiment One and Three was related to study protocol and the number of pre-trials. In Experiment One, five familiarization indentations were performed prior to testing. In Experiment Three, in addition to 5 familiarization trials, a minimum of two and a maximum of four indentations were performed prior to data collection to allow for the transition of teaching to testing of the contractions. These extra pre-trials may have allowed the spine to reach a more stable state in its viscoelasticity, such that subsequent measurements of stiffness did not change. A second difference that may explain these results is related to the time in between trials during the testing protocol. During Experiment Three, the researcher had many different tasks to perform in between indentations as compared to during Experiment One. Although every attempt was made to stay within the two minute rest period between indentations, cases did occur when, due to equipment malfunction, the rest period was slightly longer. This extra time in between indentations may have allowed the return of fluid to the tissues resulting in a similar viscoelastic condition for each indentation. In this situation, it may have been possible that two minutes between indentations was not sufficient to allow for appropriate viscoelastic reformation.

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5.C.4 Strengths and weaknesses of this study

Strengths:

This study was the first study to use an assisted, portable, stiffness device to test the effect of clinically prescribed exercises on biomechanics of the spine. Additionally, this was the first clinical use of the assisted indentor. AI demonstrated significant differences in stiffness values between conditions (which were thought to have differences based on previous research), giving support to this device's ability to quantify differences in spinal stiffness. This suggests that new outcome measures from AI are available for assessing the spine in other situations. Furthermore, knowledge from these results will provide a rationale for using this device in other situations involving the assessment of spinal biomechanics.

This study also utilized a randomized allocation method to test the differences in stiffness produced by three levels of muscle contraction: rest, hollow, and brace. This further strengthens the results of the study by eliminating the effect of order of muscular contraction. An additional strength in this study was the performance of two distinctly separate muscle contractions as evidenced by EMG and ultrasound findings. If this were not found to occur, the contractions would have been considered to be virtually synonymous, and no conclusions could be drawn regarding differences in stiffness between contractions (and most likely there would not have been any differences in stiffness, should these differences truly exist).

Lastly, reliability studies were completed for both stiffness measurements using AI and TrA contraction measurements using CSA on B-mode ultrasound images. Reliability values were excellent, increasing our confidence that the results of the study were due to true differences not due to measurement error.

Weaknesses:

The main weakness present in this study was that the contraction of the multifidus muscle was not quantified. This was not performed due to the documented inability of surface EMG to measure multifidus contraction in a valid manner.[157] Additionally, the researcher

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of this study was not trained in wire EMG and due to its invasive nature this option was not pursued. Also, B-mode ultrasound, another viable option to measure multifidus contraction, was unable to be used as the indentor was placed over the area that the ultrasound transducer would need to be located. It was determined that a contraction of multifidus was occurring in conjunction with TrA through palpation; [34] however, by not measuring multifidus function during indentation, its level of contraction throughout stiffness testing was unable to be quantified. Consequently, if the multifidus contraction was not maintained by the majority of subjects during stiffness testing, a reduced effect of the hollow on spinal stiffness could occur. This would occur without the knowledge of the researcher. Furthermore, contributions from the pelvic floor musculature, that may alter intra-abdominal pressure, cannot be excluded as contributors to the increasing spinal stiffness.

Second, another weakness exists in this study based on the limitations inherent to surface electromyography. As mentioned previously, factors such as electrode placement, skin preparation, and tension on wires could affect the validity of the sEMG output and so these were standardized within this study. However, external noise can still influence the signal and filtering was used to minimize this contamination. Lastly, many of the subjects were males of considerable strength and although all attempts were made to elicit a maximal voluntary contraction, it is possible that this was not achieved in all cases. However, the number of subjects in whom the maximal voluntary contraction may not have been achieved in comparison to the total number of subjects was relatively small, therefore, it was not anticipated that this would have a large impact on the findings of this study. Further, during an MVC contraction, motor unit firing characteristics relate to the increase in force production. The first characteristic that increases force production is increased motor unit recruitment. This can be quantified by the amplitude measurement of an EMG signal. However, the second characteristic that increases force production is the increase in firing rate of motor units. This is captured by an analysis of the frequency spectrum of an EMG signal. However, in the present study, analysis of only the amplitude of the EMG signal was used. Therefore, it must be acknowledged that the maximal voluntary contraction used in the present study was lower than that actually

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occurring as the firing rate of motor units were not taken into account. However, this was performed consistently among subjects. Therefore, although this did not confound the results of the study, the amount of muscular contraction occurring during the hollow and brace contraction may be somewhat lower than that reported.

A further limitation in this study occurred during the collection of ultrasound data of TrA. Unfortunately, an image of TrA bulk during a maximal voluntary contraction was not taken during the data collection. This prevented the study from being able to express the TrA contraction as a percentage of the maximal voluntary contraction. Additionally, this prevented the direct comparison of percentage of MVC levels between EMG findings for superficial trunk muscles and ultrasound findings for TrA contraction. Nevertheless, it has been suggested that selective isolated recruitment of TrA can occur only at low levels of contraction without subsequent contraction of the superficial abdominal muscles.[65] Therefore, in subjects with a TrA contraction exhibited on ultrasound and no contraction of the superficial muscles on sEMG analysis, it can be assumed that the TrA contraction level was relatively low.

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CHAPTER SIX
CONCLUSIONS

Overall Conclusions:

Assisted indentation was shown to have excellent intra-rater reliability as was the use of cross-sectional area measurement of transversus abdominis on B-mode ultrasound images. This suggests that these relatively new techniques have potential to be used in further biomechanical and muscle function studies, respectively. Further, when using AI to examine the effect of muscle contraction on spinal stiffness, AI was able to distinguish different conditions of stiffness within the spine. Specifically, the brace contraction provided a significantly greater stiffening affect to the lumbar spine than did the hollowing contraction. This suggests that using a general approach to muscular contraction (using all abdominal and back muscles) is more effective than using specific muscles (transversus abdominis and multifidus) to increase stiffness of the spine. However, both contractions significantly increased stiffness of the spine as compared to stiffness at rest, suggesting that the hollow and the brace contraction may be more appropriate to use in different situations requiring different levels of spinal stiffness. This knowledge may help clinicians better match exercise prescription to specific patient and situational needs.

6.A Experiment One

In summary, this study evaluated the intra-rater reliability of AI in quantifying spinal stiffness at the fourth lumbar vertebrae.

6.A.1 Conclusions

Excellent intra-rater reliability exists for AI in measuring spinal stiffness in the lumbar spine when testing male and female asymptomatic subjects with an average age of 25.3

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years and an average BMI of 22.72 kg/m^2. However, due to time-related changes in global stiffness values, it is recommended that mean maximal stiffness values should be used as the main outcome measure for AI, as they demonstrate more stability over time.

6.A.2 Relevance

Based on excellent reliability values obtained with AI, use of this device in further studies of spinal biomechanics is supported.

6.A.3. Directions for the future

Further research is necessary within the area of reliability of AI. Specifically, future studies on inter-rater reliability of this device are warranted. If good inter-rater reliability is present, the use of AI to measure spinal stiffness may be appropriate within the clinical environment. Further, AI may also have an important role in determining change in spinal stiffness in a low back pain population. Studies involving the relationship between stiffness properties of the spine and pain cessation in a low back pain population could help investigate the complex relationship between spinal biomechanics and treatment success.

6.B Experiment Two

This study was designed to evaluate the intra-rater reliability of cross-sectional area measurements of TrA contractions via B-mode ultrasound. Further, this study investigated the concurrent validity of the posterior measurement of TrA using B-mode ultrasound.

6.B.1 Conclusions

Measurement of TrA contraction on B-mode ultrasound images using a cross-sectional area technique was found to be a reliable method of quantification of TrA contraction for male and female subjects with an average age of 26 years and an average BMI of 22.7

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kg/m^2. Additionally, posterior imaging of TrA contraction (as compared to anterior imaging) was found to be a valid method to quantify TrA contraction.

6.B.2. Relevance

Due to the excellent reliability values and demonstrated concurrent validity, this technique of quantifying TrA contraction can be used in future studies.

6.B.3. Directions for the future

Further research is required to determine the inter-rater reliability of using this measurement technique in quantifying TrA contraction. As well, studies examining the reliability of TrA CSA measurement within a symptomatic population are needed. Studies comparing the responsiveness to change in TrA muscle contraction of CSA and thickness measures would be very interesting. It may be possible that due to the larger sampled area of TrA, CSA is more sensitive to small contraction levels.

6.C Experiment Three

This study determined the effect of two stabilizing contractions, the hollow contraction and the brace contraction, on lumbar spinal stability. Specifically, comparisons between stiffness measured during the rest, hollow, and brace condition were made.

6.C.1 Conclusions

On the basis of this study's findings, voluntary muscle activity from a specific trunk muscle contraction (hollow) and a general trunk muscle contraction (brace) both increase the stiffness of the lumbar spine. Specifically, the brace contraction was found to be more effective in increasing spinal stiffness in male and female subjects with an average age of 27.3 years and an average BMI of 22.79 kg/m^2.

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6.C.2 Relevance

The bracing contraction may be a more appropriate exercise to prescribe when more stiffness of the spine is needed. However, during the brace contraction, some subjects demonstrated high levels of muscular contraction, (above those recommended for the performance of these contractions), which is known to increase spinal compression. This may be an essential factor to recognize for treatment prescription of the brace contraction in regards to the importance of patient education and sufficient training to facilitate proper exercise performance and reduce the possibility of increased spinal loading.

6.C.3 Directions for the future

Further research within this area is needed to investigate if changes in the stiffness of the spine within a symptomatic population also exist. It is well established that performance of these stabilizing contractions is more difficult for patients with back pain particularly those of a chronic nature. Therefore, it is vital that these findings be established within subjects with back pain of both an acute and a chronic nature. Moreover, it is key that both different levels and different locations (such as the transverse processes of the lumbar vertebrae) be tested. It may be that the stiffness of other spinal levels are affected in a different way than the 4th lumbar vertebrae, even with the same contractions. Further, the rotational stiffness of the spine (indenting over the transverse process of a vertebra) may react in a different manner with the hollow and brace contractions than that of the spinous process. Finally, it is important to test spinal stiffness in a variety of postural positions that more closely mimic daily activities and/or pain provoking postures. In this way, more comprehensive data regarding the effect of stabilizing contractions during functional and/or painful activities could be ascertained and treatment recommendations specific to the task could be made.

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CHAPTER SEVEN

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Appendix A:
Ethical Approval from the University of Alberta Health Research Ethics Board

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213 Heritage Medical Research Centre
University of Alberta, Edmonton, Alberta T6G 2S2
p.780.492.9724 (Biomedical Panel)
p.780.492.0302 (Health Panel)
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HEALTH RESEARCH ETHICS APPROVAL

Date of HREB Meeting: February 3, 2006

Name of Applicant: Greg Kawchuk

Organization: University of Alberta

Department: Physical Therapy

Project Title: **Quantification of the effect of torso muscle contractions on lumbar spine stability as measured by ultrasonic indentation.**

The Health Research Ethics Board (HREB) has reviewed the protocol for this project and found it to be acceptable within the limitations of human experimentation. The HREB has also reviewed and approved the subject information letter and consent form.

The approval for the study as presented is valid for one year. It may be extended following completion of the yearly report form. Any proposed changes to the study must be submitted to the Health Research Ethics Board for approval. Written notification must be sent to the HREB when the project is complete or terminated.

Dr. J. Rodgers

Dr. Jennifer Rodgers
Associate Chair, Health Research Ethics Board
(B: Health Research)

FEB 13 2006
Date of Approval Release

File number: B-020206

UNIVERSITY OF ALBERTA

Capital Health

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Appendix B: Sample size calculation for Experiment One

The sample size calculation formula chosen for Experiment One was one based on the ICC statistic for measurement.

$n = v + 1$

Where:

n = number of subjects needed

v = value from a table based on Δ and power

$\Delta = (p - p_o) / (1 - p \times p_o)$

Where:

p = correlation we want to find (specified in hypothesis; ICC = 0.80)

p_o = value specified in null hypothesis (level of reliability chosen as that which would not be meaningful)

$\Delta = (0.80 - 0.50) / [1 - (0.80)(0.50)]$

$= 0.5$

When choosing a two-tailed α (2.5 %) and 80% power, $v = 22$ subjects

$n = 22 + 1$

$= 23$ subjects required

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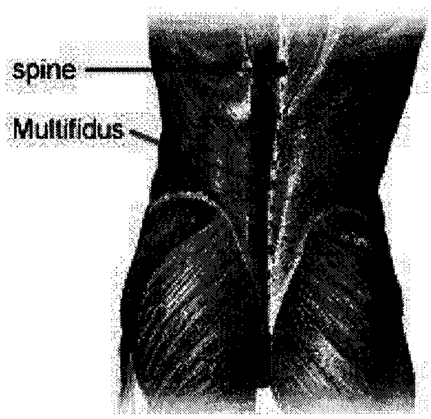
Appendix C: Recruitment poster

MALE AND FEMALE VOLUNTEERS NEEDED!!



Do you want to know more about the low back?

More about exercises that are given to people with low back pain??

You could be part of this study if:

- You are healthy
- You do not have low back pain

The study involves 1 session of about 3 hours.

If interested, please contact Tasha at

492-0563 or nliddle@ualberta.ca.

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Appendix D: Subject recruitment e-mail

Male and Female Volunteers Needed!

Want to know more about the low back? More about the exercises that are given to people with low back pain?

We are doing a study to compare how well two different exercises work to increase stability in the low back. This is being done in the Department of Physical Therapy.

You could be in this study if:

-you are between the ages of 18 - 50
-you do not have low back pain

This study will have one testing session. This will take about 3 hours.

For more information, please contact Tasha at (780) 492-0563 or nliddle@ualberta.ca.

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Appendix E: Project Information Letter

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PROJECT INFORMATION LETTER

Project Title: Quantification of the effect of torso muscle contractions on lumbar spinal stability.

Investigators:

Dr. Gregory Kawchuk, PhD, DC
Associate Professor, Department of Physical Therapy
Faculty of Rehabilitation Medicine
University of Alberta
(780) 492-6891

Ms. Tasha Liddle, BScPT
Master's Graduate Student, Department of Physical Therapy
Faculty of Rehabilitation Medicine
University of Alberta
(780) 492-0563

Purpose:
The purpose of this study is to determine how well exercises that are commonly given by health care professionals work. Specifically, we want to find out if two exercises that are given to increase stability in the back actually do.

Background:
We need to know how well two exercises that we ask patients to perform actually work. Many health care providers ask patients with low back pain to perform stabilization exercises. These exercises are thought to help give support to the low back by contracting certain muscles. In this way, the exercises are thought to help decrease back pain. We need to know which of the two exercises works better to increase stability in the spine. By knowing this, we can determine which exercise is the best for health care professionals to teach their patients. Then, the patient with low back pain will be given the exercise that supports his/her back the most. This may help decrease low back pain more quickly. It may also allow people to return to normal living more quickly.

Procedures and Risks:
If you are selected for this study, we will ask you to come to the laboratory one time. This session will last about three hours. The examination will be set up at a time convenient for you.

You will be asked general questions about yourself (age, health conditions, low back history). This helps us to describe the characteristics of our subjects. It also helps us determine who we can generalize our findings to. Your height and weight will also be measured. When measuring your height and weight, we will ask you to wear shorts (+/- bra). After this, you will be taught two exercises that are thought to stabilize the back. In order to make sure that you are contracting the right muscles, diagnostic ultrasound will be used (B-mode ultrasound). Diagnostic ultrasound has been shown to be a safe imaging method. It does not damage any tissue. Gel will be placed on your skin. This allows the ultrasound head to slide over your skin. The gel will be placed near your front hip bone. You will not experience any pain or abnormal sensations due to the ultrasound.

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During the testing session, a technology called surface electromyography (sEMG) will be used. It measures the activity of your muscles or whether they are contracting. Electrodes will be placed on your skin in five different areas. Three electrodes will be placed on areas of your stomach and two on your lower back. The skin in these areas will be shaved and rubbing alcohol will be wiped over these areas. This allows for good contact between the electrodes and your skin. This will be done prior to placing the electrodes on your skin. Then, in order to make sure we are on the right muscle, we will have you perform some trunk movements. We will provide resistance to these movements. These movements will include trunk flexion (a "sit-up"), trunk rotation, and trunk extension (bending backwards). This technology has been shown to be safe in humans. There is a small risk that your skin will be irritated after the electrodes are placed on the skin. This can occur as some people may be allergic to the gel. If this happens, it will resolve within about a week.

After the electrodes are put on, the stiffness of your back will be measured. A technique called assisted indentation will be used to measure the stiffness. A part of your back, called the spinous process, will be found by the researcher using her hands. The indentor will be placed over this area. Once the right spot is found, the researcher will push on your back with the instrument, testing your back's stiffness. At worst, you will feel mild discomfort during this part of the test. You will be given a trigger that you can squeeze if the indentation becomes painful. If you pull the trigger, the researcher will stop the indentation. The force used during the indent will stay below the level that has been determined safe in humans. The stiffness of your back will first be tested while you rest. Then you will be instructed to perform the one of the two exercises that you were taught. While you hold the muscle contraction, the stiffness of your back will be re-tested. This same procedure will be used when you perform the second exercise that you were taught.

Stiffness will be measured 4 times during rest and 2 times during each of the two exercises. This will result in a total of 10 tests of stiffness (5 measurements x 2 exercises). There is a small risk that you may develop soreness in your low back with the indentation. However, this is also temporary and will resolve within about a week.

Summary of Procedure:

Part 1:	Learn stabilization exercises B-mode ultrasound used to check that the right muscles are contracting
Part 2:	Have sEMG electrodes put on. These will be connected to a computer and your muscle activity will be measured. Have stiffness of your back tested using assisted indentation while resting (A total of 6 times) Perform each stabilization exercise Have stiffness tested during each stabilization exercise (2 measurements x 2 exercises = total of 4 times)

Reliability Study:
A small number of subjects (10) may be asked to come for one extra session to help test the reliability of our stiffness measurements. This session will take approximately 30 minutes.

Benefit:
While there are no directs benefits to you as a subject, this study's findings will help determine which of the two stabilization exercises works best to increase stability in the back. In this way, back problems can be treated more effectively. Also, this knowledge will help guide what treatments are used in clinical practice.

Risks:

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Aside from the risk of minor skin irritation and low back discomfort as mentioned above, there are no known risks associated with the methods of this study.

Confidentiality:
Only the investigators mentioned above will have access to the data of this study. All information in this study will be treated confidentially. No one will know that you have taken part in this study except the researchers. The data will be kept for at least 5 years in a secured area. It will be accessible only to the research team. Your name will not be used in any reports which are related to this study. All information in this study will be presented in a summary form.

Freedom to Withdraw:
You may withdraw from this study at any time without consequences to yourself.

If you have any questions about this study, you may contact Ms Tasha Liddle at (780) 492-0563. You may also contact Dr. Paul Hagler at (780) 492-5765. He is the Associate Dean of Graduate Studies and Research in Rehab Med. He is not involved in the study and will be willing to address any other questions or concerns.

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Appendix F: Patient Consent Form

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CONSENT FORM

Project Title: Quantification of the effect of torso muscle contractions on lumbar spinal stability as measured by assisted indentation.

Investigators:

Dr. Gregory Kawchuk, PhD, DC
Associate Professor, Department of Physical Therapy, Faculty of Rehabilitation Medicine
University of Alberta
(780) 492-6891

Ms. Tasha Liddle, BScPT
Master's Graduate Student, Department of Physical Therapy, Faculty of Rehabilitation Medicine
University of Alberta
(780) 492-0563

Do you understand that you have been asked to be in a research study?	Yes	No
Have you read and received a copy of the Project Information Sheet?	Yes	No
Do you understand the benefits and risks involved in taking part in this research study?	Yes	No
Have you had an opportunity to ask questions and discuss this study?	Yes	No
Do you understand that you are free to refuse to participate or withdraw from this study at any time? You do not have to give a reason why you want to withdraw.	Yes	No
Has the issue of confidentiality been explained to you? Do you understand who will have access to your records?	Yes	No

This study was explained to me by: ____________________

I agree to take part in this study.

______________________________ ______________________
Signature of Research Participant Date

____
Printed Name

______________________________ ______________________
Signature of Witness Date

____
Printed Name

I believe that the person signing this form understands what is involved in the study and voluntarily agrees to participate

______________________________ ______________________
Signature of Investigator Date

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Appendix G: Sample size calculation for Experiment 3

Firstly, using a study by Richardson et al (2002) [59], the relative difference of the two abdominal contractions' effectiveness at increasing stiffness was calculated (% of stiffness increase with abdominal hollowing – percentage of stiffness increase with abdominal bracing). It was found that the abdominal hollowing contraction was 17% more effective than the bracing contraction at increasing stiffness (decreasing laxity values) in the sacroiliac (SI) joint.

Using this difference in spinal stiffness (17%) known to exist between the two abdominal contraction patterns at the SI joint, it was matched to a similar level of percentage change in PA stiffness from rest calculated in a study by Shirley et al (1999) [146]. In the study, posteroanterior stiffness of the lumbar spine at L4 during varying levels of trunk extensor contractions was investigated. The percent change in stiffness that most closely matched the 17% difference was at 10% maximum voluntary contraction of trunk extensors. Using the given descriptive statistics given at the level of voluntary contraction, the standard deviation of the subjects at 10% MVC of trunk extensors was expressed as a percentage of the mean. It was calculated that the standard deviation represented 22.8% of the mean stiffness.

The current investigation is a treatment study; therefore the following formula will be used to calculate sample size:

$$\text{n/group} = \frac{2(\sigma)^2}{(\mu_2-\mu_1)^2} * f(\alpha,\beta)$$

Where:

n = sample size

σ = standard deviation

$\mu_1 - \mu_2$ = different between therapies

$f(\alpha,\beta)$ = ratio of type I and type II error to be used in the experiment.

The alpha level will be set at 0.05 and the beta level at 0.20 resulting in a power of 80%. Standard deviation and mean values were found for change in spinal stiffness between rest and a 10% MVC spinal extension contraction.

Therefore:

$$\text{n/group} = \frac{2(0.228)^2}{(0.17)^2} * 7.9$$

$$= \frac{0.10}{0.0289} * 7.9$$

$$= 3.6 * 7.9$$

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$$= 28 \text{ subjects per group}$$

Because the experimental design is patient as own control, a total of 28 subjects will be required.

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Appendix H: Screening exam and patient demographic information

Patient Name:

Age:

Gender:

Medications:

History (guiding questions):

1) Current low back pain?	Y	N
2) Any occurrences of low back pain with the last year?	Y	N
3) Lower extremity injury within the last year?	Y	N
4) Any spinal surgery?	Y	N
5) Any medical conditions? If yes, please clarify:	Y	N
6) Are you pregnant or is there any possibility that you may be pregnant?	Y	N

Height (m):

Weight (kg):

BMI:

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Appendix I: Compensation strategies

Compensation strategies that may occur during Abdominal Hollowing[34]:

1) Aberrant movement
   - -Posterior pelvic tilt
   - -Flexion of the thoracolumbar junction
   - -Rib cage depression

2) Contours of the abdominal wall
   - -No movement of lower abdomen
   - -Increased lateral diameter of the abdominal wall
   - -Visible contraction of the obliquus externus abdominis muscles fibers at their origin
   - -Patient unable voluntarily to relax the abdominal wall

3) Aberrant breathing patterns
   - -Inappropriate activation of the external obliques and internal obliques during the breathing cycle.
   - -Patient unable to perform diaphragmatic breathing pattern

4) Unwanted activity of the back extensors
   - -Co-activation of the thoracic portions of the erector spinae.

Compensations strategies that may occur during Abdominal Bracing[65]:

-aberrant breathing patterns (holding their breath)
-aberrant trunk movements (either extension or flexion of the trunk)

Compensations noted during the present study:

Abdominal hollow:

- aberrant movement (posterior pelvic tilt) and contraction of the external obliques
    - occurred in one subject out of 28 subjects

Abdominal Brace:

- aberrant trunk movements (extension of the trunk, flexion of the trunk)
    - occurred in two out of 28 subjects (one extended their trunk, one flexed their trunk)

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Appendix J: GS values – Data pooling for rest, hollow, and brace conditions. Repeated measures ANOVA and paired t-test results.

Rest Conditions

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.07	3	0.03	0.07	0.98
Rest X Gender	0.62	3	0.21	0.61	0.61
Error (rest)	26.80	78	0.34		

Table 39: A Repeated Measures ANOVA for GS Values during Rest Condition, GS measured in N/mm. Sphericity assumed.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Gender	33.05	1	33.05	5.85	0.02*
Error	146.80	26	5.65		

Table 40: A Repeated Measures ANOVA for Differences between Genders for GS Values during Rest Condition; GS measured in N/mm. Sphericity Assumed. * denotes significance at the p < 0.05 level.

(I) Gender 1	(J) Gender 2	Mean Difference (I-J)	Std. Error	Sig.	95% CI for Difference	
					Lower Bound	**Upper Bound**
1.00	2.00	1.09*	0.45	0.02	0.16	2.01
2.00	1.00	-1.09*	0.45	0.02	-2.01	-0.16

Table 41: Pair-wise comparison between different genders for GS Values during Rest Conditions; GS measured in N/mm. Sphericity Assumed. * Mean difference is significant at the p < 0.05 level.

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Hollow and Brace Conditions

	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	**Upper Bound**		
Male Hollow 1 & 2	-0.29	0.39	-1.13	0.56	-0.73	0.48
Male Brace 1 & 2	-0.30	0.94	-2.34	1.73	-0.32	0.75
Female Hollow 1 & 2	-0.31	0.40	-1.16	0.55	-0.78	0.45
Female Brace 1 & 2	0.40	1.07	-1.92	2.72	0.37	0.72

Table 42: Paired t-tests for GS values for the hollow and brace conditions; GS measured in N/mm.

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Appendix K: MMS values – Data pooling for rest, hollow, and brace conditions. Repeated measures ANOVA and paired t-test results.

Rest Conditions

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.92	3	0.31	1.49	0.22
Rest X Gender	0.92	3	0.31	1.50	0.22
Error (rest)	15.95	78	0.20		

Table 43: A Repeated Measures ANOVA for MMS during Rest Condition; MMS measured in N/mm. Sphericity assumed.

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Gender	1.82	1	1.82	0.27	0.61
Error	175.42	26	6.75		

Table 44: A Repeated Measures ANOVA for differences between genders for MMS values during the rest condition; MMS measured in N/mm.

Hollow and Brace Conditions

	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
Male Hollow 1 & 2	-0.13	0.19	-0.55	0.28	-0.68	0.51
Male Brace 1 & 2	0.95	0.71	-0.58	2.48	1.35	0.20
Female Hollow 1 & 2	-0.37	0.32	-1.06	0.32	-1.17	0.27
Female Brace 1 & 2	0.51	0.72	-1.04	2.07	0.71	0.49

Table 45: Paired t-tests for MMS values for the hollow and brace conditions; MMS measured in N/mm.

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Appendix L: Average stiffness values prior to data pooling

		GLOBAL STIFFNESS	MEAN MAXIMAL STIFFNESS
MALE			
	Hollow1	10.39 (1.84)	8.72 (1.96)
	Hollow2	10.68 (1.69)	8.85 (2.00)
	Brace1	12.16 (4.32)	10.82 (3.11)
	Brace2	12.46 (5.83)	9.87 (3.76)
	Rest1	9.17 (0.96)	7.03 (1.36)
	Rest2	9.00 (1.03)	7.38 (1.30)
	Rest3	9.01 (0.95)	7.27 (1.18)
	Rest4	8.97 (1.10)	7.01 (1.40)
FEMALE			
	Hollow1	9.75 (2.46)	8.76 (2.19)
	Hollow2	10.06 (2.08)	9.13 (1.79)
	Brace1	11.64 (5.30)	10.10 (3.60)
	Brace2	11.24 (3.00)	9.59 (3.10)
	Rest1	7.86 (1.59)	6.79 (1.33)
	Rest2	7.96 (1.45)	6.94 (1.44)
	Rest3	7.91 (1.44)	6.90 (1.52)
	Rest4	8.07 (1.68)	7.05 (1.30)

Table 46: Stiffness values (GS and MMS) for subjects during the rest, hollow, and brace conditions (prior to data pooling); measured in N/mm.

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Appendix M: EMG values – Data pooling for rest, hollow, and brace conditions. Repeated measures ANOVA and paired t-test results.

Rest Conditions

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.25	1.63	0.17	0.76	0.45
Rest X Gender	0.89	1.63	0.55	2.42	0.11
Error (rest)	9.57	42.30	0.23		

Table 47: Repeated Measures ANOVA for EMG activity of Rectus Abdominis (RA). Sphericity not assumed; Huynh-Feldt adjustment used. EMG activity (microvolts) expressed as a % of MVC.

There was a significant difference between genders for RA EMG values during the rest condition ($p = 0.05$).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.52	1.84	0.28	0.40	0.65
Rest X Gender	1.27	1.84	0.69	0.98	0.38
Error (rest)	33.70	47.83	1.30		

Table 48: Repeated Measures ANOVA for EMG activity of External Obliques (EO). Sphericity not assumed; Huynh-Feldt adjustment used. EMG activity (microvolts) expressed as a % of MVC.

No gender effects were demonstrated for the rest conditions for EO EMG values ($p = 0.26$).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	4.22	2.03	2.08	1.47	0.24
Rest X Gender	2.40	2.03	1.18	0.83	0.44
Error (rest)	74.77	52.71	1.42		

Table 49: Repeated Measure ANOVA for EMG activity of Internal Obliques (IO). Sphericity not assumed; Huynh-Feldt adjustment used. EMG activity (microvolts) expressed as a % of MVC.

No gender effects were present for the rest condition for IO EMG values ($p = 0.14$).

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Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	2.49	1.20	2.07	0.61	0.61
Rest X Gender	2.52	1.20	2.10	0.62	0.47
Error (rest)	105.78	26.00	4.07		

Table 50: Repeated Measures ANOVA for EMG activity of Thoracic Erector Spinae (T-ES). Sphericity not assumed; Huynh-Feldt adjustment used. EMG activity (microvolts) expressed as a % of MVC.

No gender effects were demonstrated for the rest condition for T-ES EMG values ($p = 0.14$).

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	4.17	3	1.39	1.67	0.18
Rest X Gender	2.28	3	0.76	0.91	0.44
Error (rest)	65.09	78	0.84		

Table 51: Repeated Measures ANOVA for EMG activity of Lumbar Erector Spinae (L-ES). Sphericity assumed. EMG activity (microvolts) expressed as a % of MVC.

No gender effects were noted for the rest condition for L-ES EMG values ($p = 0.06$).

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Hollow and Brace Conditions

	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
MALE						
Hollow 1 & 2						
RA	-0.17	0.23	-0.33	0.68	0.74	0.47
EO	-0.09	0.67	-1.53	1.35	-0.14	0.90
IO	1.96	3.10	-4.74	8.67	0.63	0.54
T-ES	-0.29	0.18	-0.67	0.09	-1.63	0.13
L-ES	0.70	0.47	-0.31	1.71	1.51	0.16
Brace 1 & 2						
RA	-0.61	1.45	-3.75	2.52	-0.42	0.68
EO	-1.90	2.93	-8.22	4.43	-0.65	0.53
IO	-2.32	7.70	-18.96	14.33	-0.30	0.77
T-ES	1.02	1.65	-2.55	4.58	0.62	0.55
L-ES	0.16	2.62	-5.50	5.82	0.06	0.95

Table 52: Paired t-test results for the comparison of contraction 1 and 2 for the hollow and brace conditions for RA, EO, IO, T-ES, and L-ES; Male results. EMG activity (microvolts) expressed as a % of MVC.

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	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
FEMALE						
Hollow 1 & 2						
RA	0.16	1.54	-0.73	1.05	0.40	0.70
EO	-1.48	1.02	-3.69	0.73	-1.45	0.17
IO	2.11	2.36	-2.99	7.20	0.89	0.39
T-ES	-0.89	0.54	-2.06	0.27	-1.66	0.12
L-ES	0.77	0.53	-0.38	1.93	1.45	0.17
Brace 1 & 2						
RA	-0.81	0.58	-2.05	0.43	-1.41	0.18
EO	0.21	2.10	-4.33	4.75	0.10	0.92
IO	-1.53	4.75	-11.80	8.73	-0.32	0.75
T-ES	2.13	1.52	-1.15	5.42	1.40	0.18
L-ES	0.74	4.15	-8.22	9.71	0.18	0.86

Table 53: Paired t-test results for the comparison of contraction 1 and 2 for the hollow and brace conditions for RA, EO, IO, T-ES, and L-ES; Female results. EMG activity (microvolts) is expressed as a % of MVC.

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Appendix N: Average EMG values (superficial muscle activity) prior to data pooling

	RA	EO	IO	T-ES	L-ES
MALE					
Hollow1	0.46 (0.81)	2.31 (2.65)	11.99 (20.91)	0.00 (0.00)	0.78 (1.68)
Hollow2	0.29 (0.60)	2.40 (2.06)	10.03 (11.96)	0.29 (0.66)	0.08 (0.30)
Brace1	2.47 (4.41)	17.53 (19.42)	34.12 (25.43)	8.34 (7.44)	20.17 (16.87)
Brace2	3.08 (3.93)	19.42 (13.05)	36.44 (24.57)	7.32 (8.09)	20.01 (15.57)
Rest1	0.28 (0.73)	0.00 (0.00)	0.20 (0.74)	0.06 (0.21)	0.82 (0.86)
Rest2	0.00 (0.00)	0.14 (0.35)	0.00 (0.00)	0.00 (0.00)	1.30 (1.28)
Rest3	0.00 (0.00)	0.00 (0.00)	0.19 (0.70)	0.22 (0.64)	1.08 (1.08)
Rest4	0.00 (0.00)	0.00 (0.00)	0.14 (0.54)	0.17 (0.45)	0.87 (1.08)
FEMALE					
Hollow1	0.78 (1.47)	2.06 (2.55)	16.96 (14.98)	0.25 (0.92)	0.99 (1.69)
Hollow2	0.61 (1.27)	3.54 (4.19)	14.85 (11.18)	1.14 (2.51)	0.22 (0.81)
Brace1	1.52 (2.02)	19.60 (10.55)	33.46 (27.16)	14.01 (15.37)	20.91 (20.45)
Brace2	2.34 (3.00)	19.39 (10.07)	35.00 (22.00)	11.88 (12.36)	20.17 (17.43)
Rest1	0.00 (0.00)	0.84 (2.07)	0.87 (2.12)	0.15 (0.40)	1.17 (1.45)
Rest2	0.00 (0.00)	0.45 (1.61)	0.13 (0.46)	0.71 (2.58)	1.14 (1.43)
Rest3	0.16 (0.56)	0.51 (1.61)	0.45 (1.61)	0.71 (2.58)	0.76 (1.21)
Rest4	0.16 (0.56)	0.77 (1.80)	0.00 (0.00)	0.13 (0.48)	0.50 (0.82)

Table 54: EMG values for trunk muscle activity during each rest, hollow, and brace condition (prior to data pooling); EMG activity (microvolts) is expressed as a % of MVC.

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Appendix O: TrA CSA measurements – Data pooling for rest, hollow, and brace conditions. Repeated measures ANOVA and paired t-test results.

Rest Conditions

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.16	3	0.05	0.67	0.57
Rest X Gender	0.06	3	0.02	0.27	0.85
Error (rest)	6.13	78	0.08		

Table 55: Repeated Measures ANOVA for TrA CSA values for the rest condition; male and female results. Sphericity assumed. TrA CSA measured in cm^2.

No gender effects were demonstrated for the rest conditions (p = 0.40).

Hollow and Brace Conditions

	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
Male Hollow 1 & 2	-0.02	0.17	-0.40	0.35	-0.14	0.89
Male Brace 1 & 2	-0.15	0.23	-0.65	0.36	-0.63	0.54
Female Hollow 1 & 2	-0.14	0.18	-0.54	0.25	-0.77	0.46
Female Brace 1 & 2	0.07	0.25	-0.47	0.60	0.27	0.79

Table 56: Paired t-test results for TrA CSA values for the hollow and brace conditions; male and female results. TrA CSA measured in cm^2.

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Appendix P: TrA mean thickness measurements – Data pooling for rest, hollow, and brace conditions. Repeated measures ANOVA and paired t-test results.

Rest Conditions

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Rest contraction	0.01	3	0.00	0.89	0.45
Rest X Gender	0.00	3	0.00	0.29	0.83
Error (rest)	0.37	78	0.01		

Table 57: Repeated Measures ANOVA for TrA mean thickness values for the rest condition; male and female. Sphericity assumed. TrA mean thickness measured in cm.

No gender differences were found for TrA mean thickness values during the rest condition ($p = 0.46$).

Hollow and Brace Conditions

	Mean	Std. Error	95% CI of the Differences		t	Sig.
			Lower Bound	Upper Bound		
Male Hollow 1 & 2	-0.01	0.04	-0.01	0.08	-0.17	0.87
Male Brace 1 & 2	-0.03	0.05	-0.14	0.08	-0.62	0.55
Female Hollow 1 & 2	-0.03	0.04	-0.12	0.07	-0.66	0.52
Female Brace 1 & 2	0.02	0.06	-0.12	0.15	0.29	0.78

Table 58: Paired t-test results for TrA mean thickness values for the hollow and brace conditions; male and female results. TrA mean thickness measured in cm.

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Appendix Q: Average ultrasound values (TrA contraction) prior to data pooling

	Contraction	Cross-sectional Area	Mean Thickness
MALE			
	Hollow1	1.84 (1.00)	0.45 (0.22)
	Hollow2	1.87 (1.22)	0.45 (0.30)
	Brace1	2.56 (1.49)	0.60 (0.31)
	Brace2	2.71 (1.44)	0.64 (0.27)
	Rest1	0.16 (0.21)	0.04 (0.05)
	Rest2	0.09 (0.37)	0.01 (0.08)
	Rest3	0.15 (0.25)	0.04 (0.07)
	Rest4	0.07 (0.21)	0.02 (0.06)
FEMALE			
	Hollow1	1.72 (0.72)	0.42 (0.17)
	Hollow2	1.86 (0.70)	0.45 (0.17)
	Brace1	1.89 (1.26)	0.45 (0.26)
	Brace2	1.82 (1.17)	0.43 (0.23)
	Rest1	0.03 (0.35)	0.01 (0.09)
	Rest2	0.07 (0.22)	0.02 (0.05)
	Rest3	0.15 (0.39)	0.04 (0.10)
	Rest4	0.02 (0.26)	0.00 (0.06)

Table 59: Cross-sectional area (cm^2) and mean thickness (cm) for transversus abdominis prior to data pooling.

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