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THE EFFECT OF VARIATIONS IN THE STRENGTH AND TYPE OF SPINAL MUSCLES ON THE STABILIZATION OF THE LUMBAR SPINE VIA FOLLOWER COMPRESSIVE LOAD MECHANISM

by

Ino Song

A thesis submitted in partial fulfillment
of the requirements for the Master of Science
degree in Biomedical Engineering in the
Graduate College of
The University of Iowa

May 2017

Thesis Supervisor: Professor Tae-Hong Lim

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CERTIFICATE OF APPROVAL

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MASTER'S THESIS

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This is to certify that the Master's thesis of

Ino Song

has been approved by the Examining Committee for
the thesis requirement for the Master of Science degree
in Biomedical Engineering at the May 2017 graduation.

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To my family

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ABSTRACT

The ligamentous human lumbar spine is considered as a long and slender column, which can be buckled when subjected to the axial compressive load even less than 100N. However, previous *in vivo* study showed that the compressive force acting on the spine predicted by intradiscal pressure was exceed 2600N. Meanwhile, recent experiments suggested that, when the compressive force is subjected to the lumbar spine along the spinal curvature (follower load), the lumbar spine may support up to a compressive force of 1200N without buckling while maintaining its flexibility. Since such a follower load is directed tangential to the curved column over the entire length, the lumbar spine subjected to a follower load should experience only pure compressive force components with zero shear force components.

It is generally agreed that the ligamentous lumbar spine can be stabilized by applying the muscle forces (MFs) *in vivo* creating follower compressive loads (FCLs). In previous studies, computational model of the lumbar spine showed the feasibility for spinal muscles to stabilize the lumbar spine via the FCL mechanism, which supports the hypothesis of FCLs as normal physiological loads in the spine *in-vivo.* In addition, the muscle forces of short intrinsic muscles (SIMs), such as interspinales, intertransversarii, and rotatores may increase the stability of the lumbar spine (i.e., deflection of the spinal column or trunk sway) significantly. However, the mechanical roles of SIMs for spinal stability have not been quantified and understood well.

A finite element (FE) model with optimization model of the lumbar spinal system was used in this study. Both models were consisted of 122 pairs of spinal muscle fascicles including 54 SIMs fascicles. The variation of spinal muscle strength was

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simulated by changing the values of MFCs of long muscles as well as SIMS from zero to 90 N/cm^2. Five different MFC conditions of both long muscles and SIMs in the spinal system were investigated in five different postures, which are neutral standing, flexion 40°, extension 5°, left axial rotation 10°, and right lateral bending 30°. The trunk displacement (TD) and joint loads including joint reaction forces (JRFs) and moments (JRMs) predicted from 25 cases of MFC variation were compared in order to investigate the effect of the strength of spinal muscles on the stabilization of the lumbar spine in a given posture.

The results showed that small trunk sways (< 2mm) were predicted when MFCs of both long muscles and SIMs were average or higher regardless of the spinal postures. In contrast, no optimum solution or unstable conditions were predicted in many cases of the weakening of the long muscles, especially in flexion and lateral bending postures. Although the FCLs were created in most of the cases regardless of MFC-S when working with strong long muscles, higher joint loads were predicted as a result of weakening of SIMs. In addition, even if the long muscles were strong, absence of SIMs induced spine buckling in some cases of extension and axial rotation postures.

The results from this study imply that although the effect of MFCs variation of long muscle and/or SIMs was varied depending the spinal postures, the simultaneous use of both SIMs and long muscles is necessary for stabilization of the spine in any physiological posture with minimum joint loads for maximum safety.

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PUBLIC ABSTRACT

Low back pain (LBP) is one of the most prevalent afflictions, which cause billions dollars of healthcare cost in each year. Even though many clinical strategies for LBP treatment have been suggested, current understandings of the basis of LBP problems are limited and insufficient. Out of possible causes of LBP under consideration, the basis of the problem commonly agreed in the field of spine research is the mechanical insufficiency of the spinal column, which is known too flexible to support the upper body weight.

Previously, it was suggested that the normal spinal load could be the compressive forces whose direction is parallel to the curvature of the lumbar spine while the shear forces whose direction is perpendicular to the spinal curvature are abnormal forces. Although those biomechanical loads on the spine are known to be closely associated with the spinal muscle control system, the significant role of back muscles, especially the short intrinsic muscles (SIMs), has not been studied sufficiently. For these reasons, in this study, the effect of variations in the strength and type of spinal muscles for spinal stabilization was investigated in various postures using mathematical and computational methods.

Throughout this study, it was found that the variation of maximum force capacity of spinal muscles affected to the spinal stability as well as the joint loads on the spine, although the magnitude of the effect was varied depending on the muscle types (Long muscles and SIMs) or the spinal postures. Therefore, it can be concluded that the concurrent use of both SIMs and long muscles is necessary for spine stabilization in any physiological posture with minimum joint loads for maximum safety.

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LIST OF ABBREVIATIONS

ALL - Anterior Longitudinal Ligament

AR - Axial Rotation

CG - Center of Gravity

EMG - Electromyography

EO - External Oblique

ES - Erector Spinae

EXT - Extension

FCLs - Follower Compressive Loads

FE - Finite Element

FLP - Follower load path

FLX - Flexion

FSU - Functional Spinal Unit

GC - Geometrical Center

IO - Internal Oblique

JRFs - Joint Reaction Forces

JRMs - Joint Reaction Moments

LB - Lateral Bending

LBP - Low Back Pain

LD - Latissimus Dorsi

MFs - Muscle Forces

MFC-L - Muscle Force Capacity of Long muscles

MFC-S - Muscle Force Capacity of Short intrinsic muscles

NS - Neutral Standing

PLL - Posterior Longitudinal Ligament

PM - Psoas Major

QL - Quadratus Lumborum

RA - Rectus Abdominis

ROM - Range of Motion

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SIMs - Short Intrinsic Muscles

SPI - Serratus Posterior Inferior

TDs - Trunk displacements

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CHAPTER 1

INTRODUCTION

Low back pain (LBP) is one of the most prevalent affliction. More than 75% of people have LBP in their lifetimes [1], and the United States is spending more than 120 billion dollars a year for the treatment of LBP [1-4] The cause of LBP, however, still remains unclear. Out of possible causes of LBP under consideration, the basis of the problem commonly agreed in the field of spine research is the mechanical insufficiency of the spinal column which is known too flexible to support the upper body weight [5, 6].

In 1744, Euler proposed the theory of critical load of a slender column. Critical load is the maximum compressive load that can be applied to a column without buckling. Based on this theory, Crisco et al. proved that the compressive load on spinal column is about 88N [5, 6]. However, Nachemson measured in-vivo that the compressive pressure on the intervertebral disc (IVD) may exceed 2600N, which is much higher than the results from Crisco [7]. Meanwhile recent experiments done by Patwardhan et al. revealed that, when the compressive force is applied along the curvature of the lumbar spine (follower load), the lumbar spine may support up to a compressive force of 1200N without buckling while maintaining its flexibility [8]. Since such a follower load is directed tangential to the curved column over the entire length, the lumbar spine subjected to a follower load should experience only pure compressive force components with zero shear force components. Such reasoning made Patwardhan et al. propose a

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hypothesis that a compressive force may be a normal physiological load in the lumbar spine whereas a shear force in the direction transverse to the spinal curve is abnormal. The application of the follower load has been used as a standard method to apply physiological compressive loads on the lumbar spine during various in-vitro biomechanical tests of the lumbar spine over a decade.

Meanwhile, investigators in the Spine Biomechanics Laboratory of the Department of Biomedical Engineering in the University of Iowa have been testing Patwardhan et al's hypothesis experimentally and computationally. Kim et al. conducted in-vivo experiments using rats and observe detrimental effects of the shear force on the rat lumbar, such as the development of degeneration in the disc subjected to a shear force and pain behavior [9]. DuBois confirmed such detrimental effects and also found that the application of shear force on the L5-L6 level longer than 4 weeks could cause a scoliosis like deformity in the thoracolumbar spine in the rat [10]. The results of these in-vivo rat experiments support Patwardhan et al's hypothesis at least in part.

Another series of studies done in the Spine Biomechanics Laboratory were computational analyses. Han et al. formulated a computational model of the lumbar spinal system including 232 spinal muscle fascicles and conducted optimization analyses [11]. Their model predictions demonstrated the feasibility of spinal muscle contraction patterns that create the follower compressive loads (FCLs) in the lumbar spine in neutral, flexed, and extended posture (sagittal postures). Kim et al. developed finite element (FE) models of the lumbar spinal system whose geometric structures were matched with those of Han et al's optimization model. The results of their FE analyses using muscle forces (i.e., optimum solutions predicted from Han's analyses) as input data clearly showed that

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there exist FCL creating spinal muscle forces (MFs) which can stabilize the lumbar spine in all sagittal postures. Then, Wang extended these computational analyses and was able to find spinal MFs which can stabilize the lumbar spine in 3-D postures (neutral posture, flexion 40°, extension 5°, lateral bending 30°, and axial rotation 10°) while creating FCLs in the lumbar spine [12]. The results of these studies suggest that there exist numerous spinal MF combinations creating FCLs in the lumbar spine while the spine can be in a stable condition, which supports the hypothesis of FCLs as normal physiological loads in the spine in-vivo. In addition, careful analyses of the results also revealed that the magnitudes of FCLs and the stability of the lumbar spine (i.e., deflection of the spinal column or trunk sway) vary more sensitively to the force changes in short intrinsic muscles (SIMs) than to those in long spinal muscles.

Although not investigated objectively, the stabilizing role of SIMs, such as interspinales, intertransversarii and rotatores, has been suggested in the past. Bergmark identified SIMs as “local stabilizing system” based on the findings that SIMs provided an increased segmental stiffness to maintain the stability of spine while there was upper limit on the possible activation of the long and large spinal MFs [13]. Cholewicki and McGill also suggested that increasing of activities of SIMs prevent spine instability [14]. Mechanically, the direct attachment of SIMs to the vertebrae may allow SIM forces to change the direction of the internal force in each segment and to achieve effective resistance to bending and/or axial rotation of individual vertebra although SIMs can generate much less force and their moment arm length are shorter than long muscles. Plenty of muscle spindles attached to intertransversarii and rotatores muscles indicates that SIMs contributes to the rapid reaction to the external perturbations [15]. Such

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mechanical contribution of SIMs may attribute to minimum deflection of the spine required for maximum stability. It was also predicted from Wang's study that the lumbar spine could be stabilized by long spinal muscles only (i.e., without contribution of SIM forces) but with costs of significant increases in the trunk sway and internal loads in all lumbar segments [12]. Biological observations revealed that the primary muscle fiber of SIMs is type I fibers, which have high fatigue resistance, whereas long muscles consist majorly of type II fibers which have low fatigue resistance [16, 17]. Higher fatigue resistance expected in SIMs should be physiologically more advantageous to keep the upright postures for long time during daily activities. As such, all of the postulations, observations, and model predictions reported in the literature indicate that the contribution of SIMs may be necessary to stabilize the spine effectively.

However, the mechanical roles of SIMs in the lumbar spine have not been quantified and understood sufficiently. The purpose of this study was to investigate the stabilizing roles of SIMs quantitatively using computational optimization and FE analyses. Hence, magnitudes of trunk displacements (TDs) and joint loads (joint reaction forces and moments) of the lumbar spine were compared when the maximum muscle force capacity (MFC) of long and/or short intrinsic muscles were varied while creating FCLs.

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CHAPTER 2

LITERATURE REIVEW

Low back pain (LBP) is one of the most common ailments that result in the significantly negative impact on the quality of life as well as on the socioeconomic. LBP is the major cause of hours lost at work and attributes to billions of health care dollars in modern society. It also has been identified as second most frequent reason for visits to physicians, the fifth ranked cause of admissions to hospitals and the third most common reason for surgical procedures. Approximately one percent of the U.S. population is chronically disabled because of back pain and an additional one percent is temporarily disabled [4]. The 1992 to 1994 National Health Interview Surveys report that back pain resulted in an average of 297 million restrictive-activity days per year and 87 million bed-disability days.

Unfortunately, the recent advances in treatment modalities and new findings from enormous research efforts were not good enough to address LBP problems yet. Specific causes of LBP remain largely unknown. A myriad of therapeutic procedures (i.e., fusion devices and intervertebral joint preservation technologies) have been introduced but their clinical effectiveness is controversial in many cases [18]. In contrast, there is reasonable evidence that LBP is improved by physical therapy treatment such as exercise, suggesting the importance of postural muscle control to improve the stability of the lumbar spine [19-22]. Although still unclear, it is generally agreed that a substantial portion of the LBP problem results from mechanical imbalance between the unstable spinal column and the inadequately controlled spinal muscle forces (MFs), which is often called as spinal instability.

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The concept of spinal instability, however, remains uncertain and controversial because of numerous limitations extremely difficult to overcome. For example, large subject-to-subject variations in their anatomic conditions, voluntary efforts to produce spinal motion, the presence of muscle spasm and pain, have produced uncertainties in testing the hypothesis about spinal instability. Other reasons for the uncertainties would include the limited accuracy of methods for spinal motion measurement, too many spinal muscles for experimental simulation, and the lack of knowledge of spinal muscle control mechanism. Various computational methods were developed in order to address these difficulties of experimental tests of the hypothesis and have been used to investigate either independently or in combination with the experiments. The results of these computational studies were helpful for better understanding of the stabilization of the spinal column. Yet, current understandings of biomechanical origin of LBP problems are largely limited and insufficient. Further efforts have to be made to address the biomechanical uncertainties relationship between the spinal instability and LBP problems.

2.1 Spinal Stabilization System

It is generally agreed that an inherently flexible and unstable spinal column can be stabilized by the additional forces produced by the precise activation of surrounding muscles (so-called spinal (or back) muscles) in a controlled manner. So, the spinal stabilization system consists of three subsystems: (1) spinal column, (2) spinal muscles, and (3) motor control unit [23].

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Spinal Column: The spine is a long and slender column consisting of 26 bones (vertebrae) connected with each other by the intervertebral discs and numerous spinal ligaments. In a frontal view, the ligamentous spinal column generally appears straight and symmetrical with respect to the mid-sagittal plane whereas it has three curves regions in a lateral view. Two curves are convex anteriorly in the cervical (top) and lumbar (bottom) regions while convex posteriorly in the thoracic (middle) region. The cervical and the lumbar region of the spine consist of 7 cervical vertebrae and 5 lumbar vertebrae, respectively. The thoracic region of the spine is made of 12 thoracic vertebrae where the rib bones are connected. Out of these 3 regions, the lumbar region, called lumbar spine, is known to play a major role in supporting the upper body weights while allowing the motion the upper body during normal everyday activities. As a result, anatomic structures above the lumbar spine (such as, head, cervical and thoracic spines, rib cage, and upper extremities, etc.) have been simulated as one rigid body, called “trunk” in biomechanical studies. The trunk weight includes the weights of all body parts above the sacrum. The position of the trunk center of gravity (TCG) varies widely in the literature while it is generally accepted that the trunk weight produces flexion moment to the lumbar spine in the neutral standing posture.

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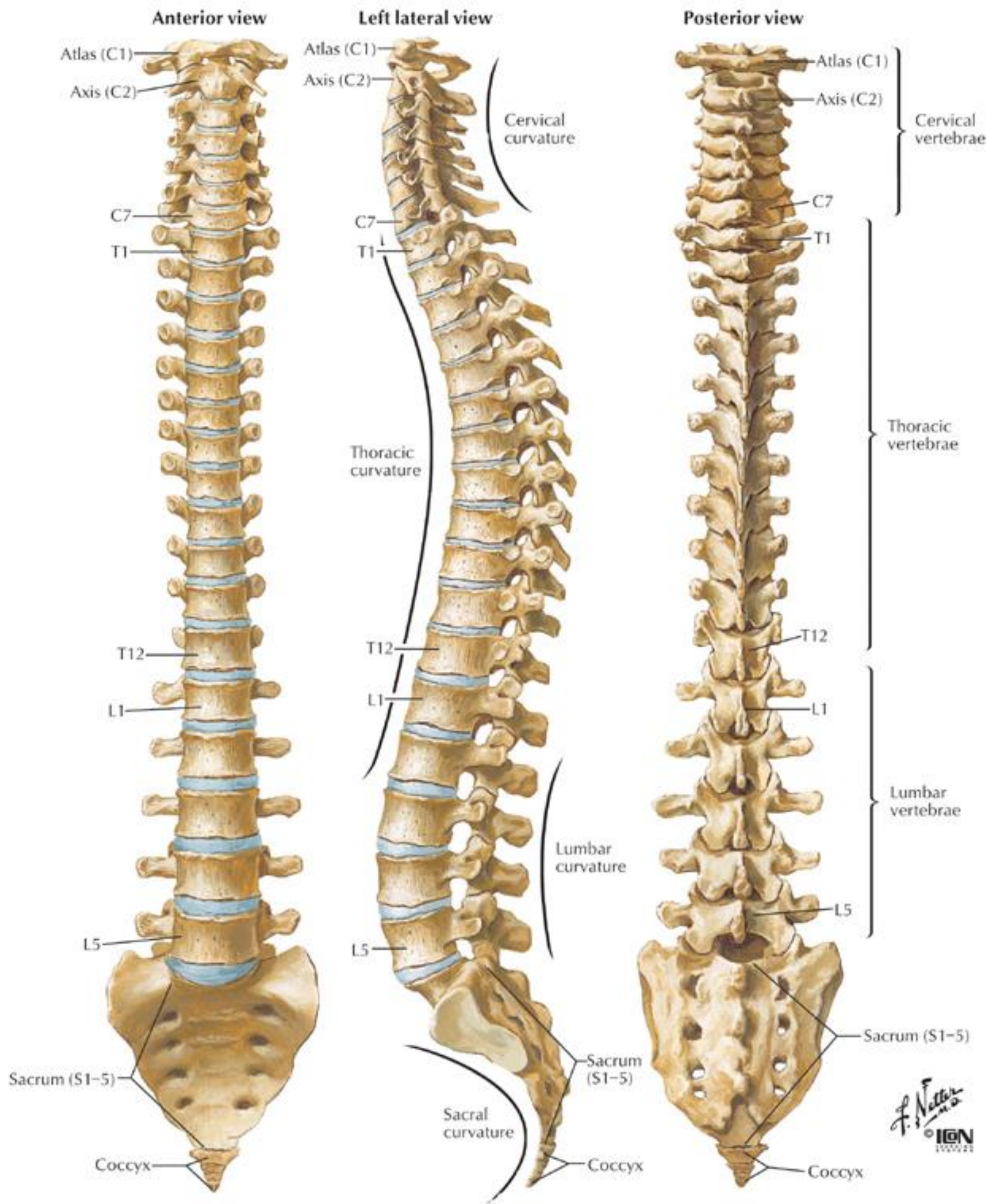

Figure 2-1 The entire human spine in anterior, lateral and posterior views.

Source: http://www.backpain-guide.com

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The lumbar spine consists of 5 lumbar vertebrae (L1 to L5). The most superior vertebra (L1) is connected to the trunk through the intervertebral joint between L1 and the most inferior thoracic vertebra (T12), which is also called thoracolumbar junction. The most inferior lumbar vertebra (L5) is connected to the sacrum (Sa) through the intervertebral joint. The unit made of two adjacent vertebrae connected by one intervertebral joint is called a functional spinal unit (FSU) or more frequently a motion segment.

The intervertebral joint consists of the intervertebral disc (IVD), two facet joints, and 7 ligaments. The IVD is located between the adjacent vertebral bodies (anterior portion of a vertebra with respect to the spinal canal). All IVDs in the lumbar region is thicker anteriorly, which contributes to the lordosis of the lumbar spine. The IVD has a complex anatomic structure and undergoes the degenerative changes with aging. Its degeneration is believed to be closely associated with LBP problems and has been attracting the attention of researchers in a variety of expertise. Nonetheless, biomechanically, the IVD is a major load-bearing structure as well as a flexible structure providing the physiological movements of the trunk and has been simulated as a nonlinearly elastic disc in numerous finite element (FE) analyses.

The intervertebral joint has two facet joints located at the right and the left posterolateral aspect of the motion segment. Facet joints are true synovial articulations enclosed in capsular ligament. They are one of the main structures for the stability of the motion segment although their load-bearing role is known much less significant than that of the IVD. Biomechanically the facet joint has been simulated as a simple structure

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(either a flexible rod or a spring) for macro-scale spine biomechanical analyses [11, 12, 24].

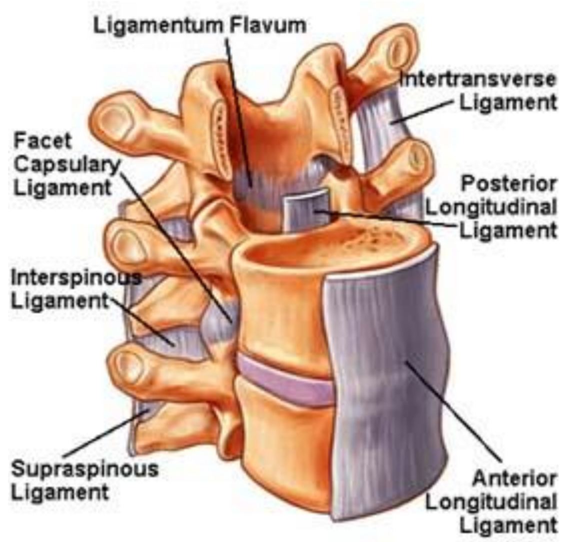

Figure 2-2 Spinal ligaments in a motion segment.

Source: https://www.spineuniverse.com

A motion segment has seven different ligaments, anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum (LF), intertransverse ligament (ITL), capsular ligament (CL), interspinous ligament (ISL) and supraspinous ligament (SSL). Mechanically, they are uni-axial structures subject to only tensile forces, and their mechanical role is to prevent the excessive intervertebral motion. Such roles of the spinal ligaments have been simulated separately using a tension-only spring element in FE analyses. However, they also can be incorporated adequately into a computational model with more simplification. For example, Kim BS et al. considered the ALL and PLL as a part of the IVD because they are firmly attached to the IVD and included their

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stabilizing roles into the elastic properties of the IVD. The stabilizing role of the CL was included as a part of an elastic rod simulating the facet joint. Furthermore, the mechanical contribution of the other ligaments was modeled using a tension-only spring element located between the adjacent spinous processes. An FE model of the ligamentous lumbar spinal column with such simplifications were found to demonstrate segmental motions similar to those found in previous in-vitro biomechanical tests [24].

Spinal Muscles: Numerous spinal muscles have been identified working in the lumbar region directly or indirectly. Their anatomical structure is extremely complex but can be organized into three groups according to their location: (1) superficial muscles; (2) intermediate muscle; and (3) deep muscles as shown in Figures 2-3.

The superficial muscles include latissimus dorsi (LD), external oblique (EO), internal oblique (IO), and rectus abdominis (RA). LD is a flat muscle of triangular shape covering the lumbar and lower half of the thoracic region. It is located posteriorly to the spine and also called a back muscle. EO is located most superficially in the anterior-lateral abdomen while IO lies just underneath the EO. The RA is long and flat and forms the front of the abdomen wall. EO, IO and RA are also called abdominal muscles.

The intermediate muscles are erector spinae (ES) and serratus posterior inferior (SPI). The ES is the largest back muscle group consisting of three sub-groups in the lumbar region; illiocostalis; longissimus; and spinalis. The most lateral portion of the ES is the illiocostalis muscles, which arise from the tip of the spinous processes. The middle portion of the ES is occupied by the longissimus muscles (also called as longissimus thoracis) consisting of numerous fascicles. These fascicles arise from the sacrum, spinous processes of the lumbar vertebrae and transverse processes of the last thoracic

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vertebra and attach to the transverse processes of the lumbar vertebrae, ES aponeurosis, ribs, and costal processes of the thoracic vertebrae. The medial portion of the ES is spinalis muscles, which have three parts: spinalis thoracis, spinalis cervicis and spinalis capitis. The spinalis thoracis arises from the spinous process of L3 – T10 and inserts in the spinous process of T8-T2.

The deep layer muscles include rotatores, intertransversarii, interspinales, quadratus lumborum (QL), psoas major (PM) and multifidi (or multifidus muscles). The multifidus is identified most developed in the lumbar region. It is located just superficially to the spinal column and spans three motion segments. It arises from the back of the sacrum, aponeurosis of origin of the sacrospinalis, medial surface of the posterior iliac spine and the posterior sacroiliac ligaments in the sacral region as well as from all the mammillary processes in the lumbar region. In the thoracic region, the multifidus arises from all the transverse processes. The QL is the deepest abdominal muscle positioning on the posterior abdominal wall. It arises from aponeurotic fibers into the iliolumbar ligament and the internal lip of the iliac crest and inserts to the lower border of the last rib and to the apices of the transverse processes of the upper four lumbar vertebrae. The psoas major (PM) is located one the anterior lateral side of the vertebral column and brim of the lesser pelvis. It is divided into a superficial and deep part. The deep part originates from the transverse processes of L1 – L5. The superficial part originates from the lateral surfaces of the last thoracic vertebra, lumbar vertebrae 1 to 4, and from neighboring intervertebral discs. The rotatores, interspinales, and intertransversarii are deepest and shortest muscles in the lumbar region. These muscles are located beneath the multifidus and span over one or two segment. Because of these

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reasons, these muscles are also call as short intrinsic muscles (SIMs). These muscles exist in pairs on both sides and are known to play the main role in the slight adjustment of segmental motion and the stabilization of adjacent levels [25]. These SIMs, however, are mostly overlooked in spine biomechanics analyses.

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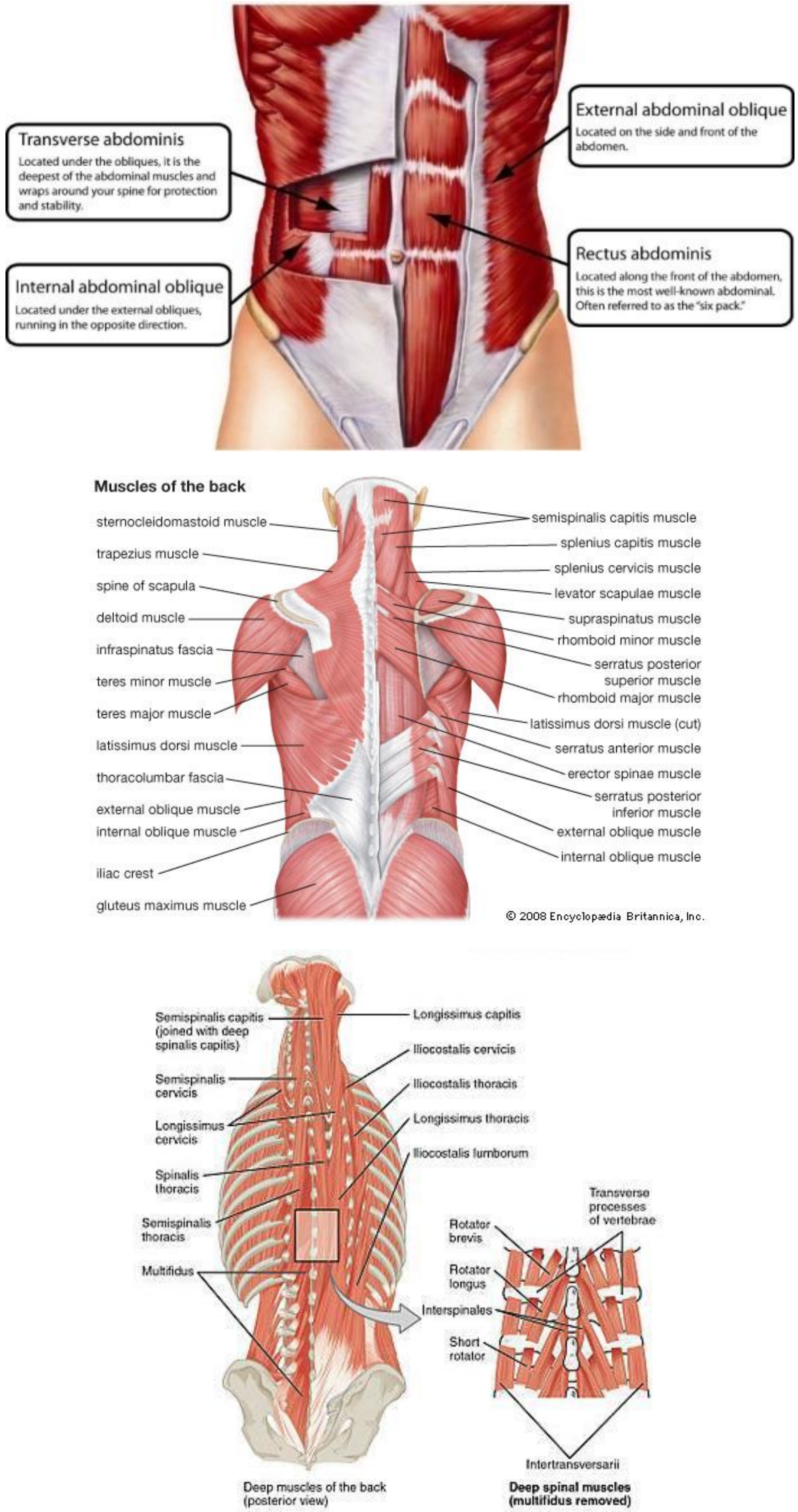

Figure 2-3 Illustration of the spinal muscles at different layers and a view of the muscle orientation [12].

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As described briefly, spinal muscles are multifarious and their anatomical arrangement is highly intricate. Each muscle can do its own functions depending upon its orientation and location while contributing to the biomechanical functions of the spinal muscles: (1) the stabilization of the lumbar spine and (2) generation of the upper body movement during daily activities. For example, the ES muscles are identified as back extensors. Spinal muscles need to produce forces to do such biomechanical functions. Although not separately investigated, it is reasonable to expect that the spinal muscles have similar biomechanical characteristics to other skeletal muscles, which are known to produce forces actively and/or passively. The active force results from the contraction of a muscle while the passive force results from the elastic property of the muscles like a tension spring force during its elongation. The spinal MFs predicted in this study as well as in previous studies indicate the total force in each muscle required to perform a given function and this include both active and passive forces [11, 12, 24].

Motor Control Unit: According to Panjabi's postulation, in-vivo the spinal column provides information of the position, motion, and loads to the central nervous system which transforms such information into action in terms of force generation in spinal muscles [26]. This hypothesis is reasonable and well accepted in the field of spine research. However, there are no studies testing the hypothesis, and the mechanism of in-vivo controlling the spinal muscle forces required for the biomechanical functions of the spine remains unknown. Nevertheless, it is reasonable to think that the motor control unit in the spinal stabilizing system should be a part of the central nervous system.

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2.2 Stability and Flexibility of the Lumbar Spine

The biomechanics of the lumbar motion segment have been intensely investigated and well characterized in terms of its range of motion (ROM) and/or stiffness (or flexibility). A typical load-displacement relationship of the motion segment is non-linear (Figure 2-4), demonstrating the increase in the slope of the load-displacement curve (i.e., the stiffness), which cannot be represented by a single stiffness value. Panjabi suggested the use of two parameters: neutral zone and range of motion (ROM). The neutral zone is a part of ROM within which the slope of the curve is small (i.e., low stiffness or high flexibility zone). The ROM can be considered equivalent to the proportional limit in a conventional test for the measurement of material properties and indicates the maximum motion without permanent deformation or structural injuries in the motion segment resulting from an applied load. Average ROMs of the intact lumbar segment in response to applied moment of 8 Nm in Abumi et al's experiment were about: 8° in flexion; 4° in extension; 3.5° in one side axial rotation; and 5.5° in one side lateral bending [27]. The axial stiffness of the lumbar segment in response to a compressive force was about 1600 N/mm in Li et al's study [28]. In addition, Adams et also showed that lumbar motion segments could withstand axial compression force 3000 – 5000 N without damage or buckling [29]. As such, a lumbar motion segment is not only strong and stable enough to support a high loads without failure but also adequately flexible to allow 3-dimensional segmental motions which may be necessary for effective movement of the trunk in-vivo.

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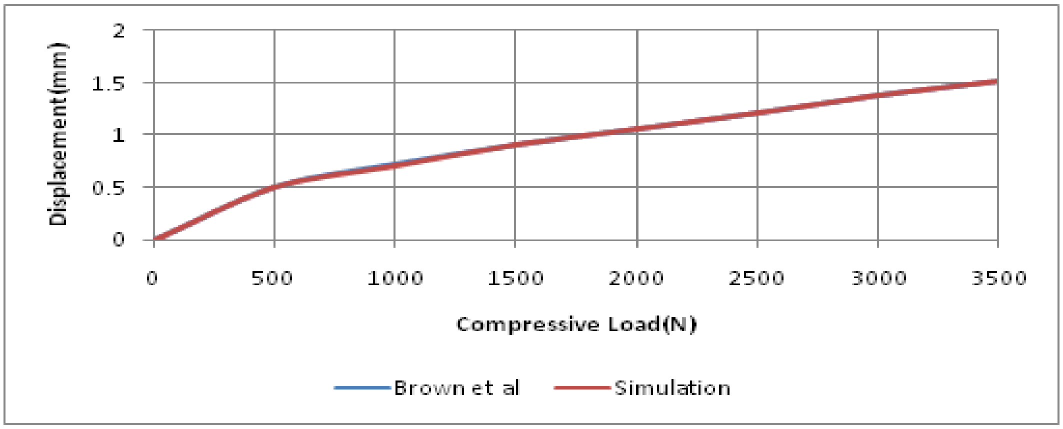

Figure 2-4 Comparison of compressive load vs. axial displacement between experiment and simulation [24].

The vertical stacking of 3 lumbar segments produced a lumbar spinal column with a lordotic curve. The segmental flexibility produces the deflection of the spinal column, which allows the trunk movement. The average range of the trunk movement, in other words the deflection angles of the lumbar spine in terms of Cobb's angle, measured in-vivo were: approximately 58.6 degrees between T12 and S1, but makes the spinal column unstable (or too flexible) to support the trunk weight without buckling [30]. Crisco et al. observed the buckling behavior of the lumbar spine resulting from the application of the vertical load less than 100 N on the L1 vertebra [5, 6]. However, Nachemson estimated the compressive load on the lumbar spine by measuring the disc pressure on the L3 disc, which can be as high as 2100N [7]. Although exposing repetitive excessive load (2000N) on the spine could result in the physiological problem on the spine as well as the disc, ligamentous lumbar segments are known to be stable under the load greater than 100N without permanent deflection of the spine. Since the spinal musculature is the only active source which can generate the force on the ligamentous spine *in vivo* situation,

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understanding the muscle forces mechanism is needed to know how the stability and the flexibility of the spine can be maintained.

2.3 Follower load on the spine

In the theory of elastic stability, which introduced in the work of Euler, applying an axial compressive load on the long and slender structures may induce the buckling or permanent deflection of the column. The axial force that results in the structural alternation from the stable to unstable condition (i.e., smallest inclination or deflection of the column) is called a critical load. This critical load on the structure can be changed by the variation of many factors, such as the higher stiffness, shorter length, lateral support, or the direction of the load. Regarding the spinal structure, although the length and rigidity of the spine is pre-determined, the load direction of the spine can be decided by the spinal muscle contraction. Therefore, the compressive load applied to the spine is the only modifiable factor for increasing critical load as well as the stability of the spine, which named a follower compressive load (FCL).

As shown in figure 2-5, the concept of the follower load was introduced by Timoshenko, Gere and Bazant [31, 32]. They defined the follower load as a force of which direction is tangential to the deflection curve of the beam. Then, with respect to the small disturbance, Bolotin proved that the critical load on the column under follower compressive load is much higher compared to the column under axial compressive load [33]. Therefore, if the follower load represents the potential physiological load on the spine, it may explain how the spinal column can support the large compressive load while maintaining its flexibility.

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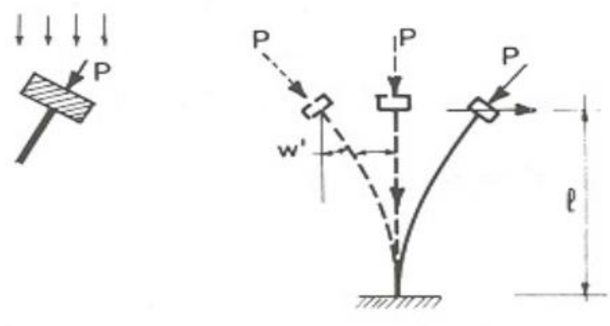

Figure 2-5 Beam loaded by a constant follower force [32].

In spine research field, Aspden first introduced the concept of compressive load traveling along the curvature of in spine column. It is generally accepted that, the joint forces in the spinal column can be classified to axial compressive force and transverse force (or shear force). He proved that when the force line exists inside of the spine in order for stabilization, axial compressive force is the only force that can be existed [34]. In-*vitro* study done by Farfan at el. also showed that there was no obvious injury resulted from a large axial compressive load on the intervertebral joint [35]. It implied that applying axial compressive load to joint might not induce instability of the spine. Then, Patwardhan suggested new *in vitro* experimental method to apply compressive follower load to ligamentous lumbar spine using loading cables (Figure 2-6). The cable was located approximately at the center of rotation of the vertebrae, and it determined the load path to tangent to the spinal curve while minimizing the shear force. They found that spine under follower load increased the ability to sustain the compressive load up to 1200N without buckling while maintaining its flexibility (i.e., minimized angular changes under load) [8]. The limitations of this study were that because of the cable placement

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and orientation, the test was successfully conducted only in sagittal plane, and the applied load cannot be increased greater than 1200N. In addition, although they showed that the follower load might be the potential mechanism of the spinal stabilization, the spinal muscles, which create the follower load on the spine, were not determined in the series of Patwardhan et al's studies of the follower load.

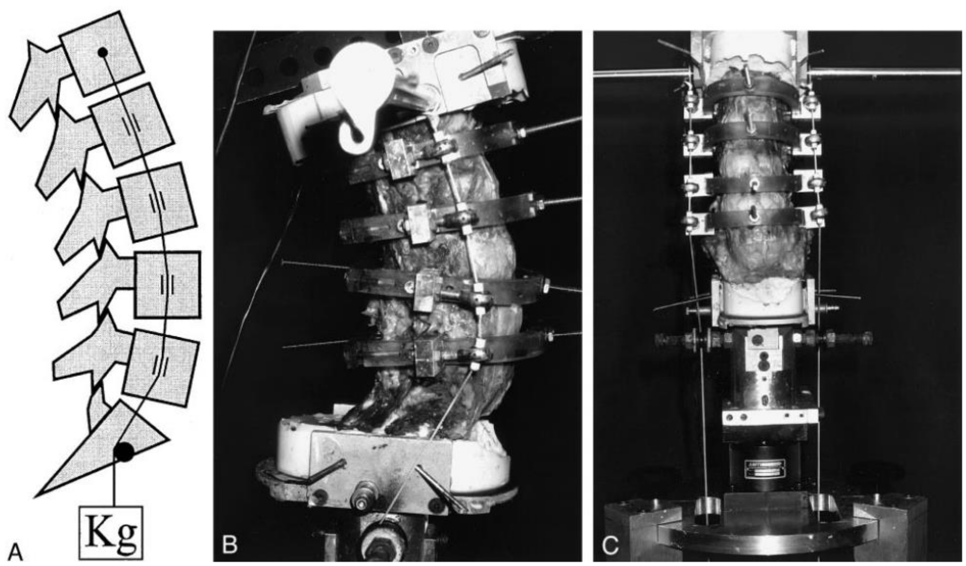

Figure 2-6 A human cadaveric lumbar spine subjected to a compressive follower load [8].

There was an effort to determine the spinal muscles creating a follower compressive load in the lumbar spine using a 3-D FE model of the lumbar spine including 117 pairs of trunk muscles (58 pairs of superficial muscles and 59 pairs of deep muscles) in a static manner [36]. Their model was developed from T12 to S1, and the squared sum of all of the resultant joint forces and moments were minimized using optimization technique. Their results showed that the posture creation and follower forces could be well preserved regardless of the MFCs while shear forces were varied significantly when the MFs of superficial muscles were restricted by 25%. However,

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they failed in finding the active muscle forces combination for creating the perfect follower load because non-zero shear forces were predicted in all cases and called it as a modified follower load.

On the other hand, Han et al. suggested the optimization model of the lumbar spine under follower load, which was considered as the potential mechanism of the spine [37]. His model included 116 pairs of spinal muscles, and 27 pairs of them were the short intrinsic muscles (SIMs). They used an optimization technique to solve the muscle forces and joint loads. Their results showed that the perfect follower compressive load without shear force could be created by using combination of active MFs when the external load was applied, and no shear and bending moment indicated vertebrae is not rotated so that it can provide the maximum stability of the lumbar spine with minimum MFs. However, the limitations of this study were that they showed the results in the sagittal plane postures, and the stability of the spine under follower load did not quantified well.

Kim et al. used Han's optimization model to solve the muscle forces creating FCLs on the spine and built FE model of the lumbar spine in order to investigate the stability of the spine [24]. Their FE model showed that the stability (i.e., the permanent deformation) of the spinal column when MFs provided from optimization model were applied. The validation of the FE model for the reliability was confirmed by comparing the range of motion of complete lumbar spine as well as the segmental range of motions in all lumbar levels. They showed the feasibility of the MFs creating pure FCLs on the lumbar spine in neutral standing posture as well as flexion and extension postures. In

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addition, they found that FCLs created only vicinity of the base line curve would provide sufficient stability of the spine.

Following Kim et al.'s research, Wang founded the feasibility and stability of the lumbar spine in extended postures including axial rotation and lateral bending postures. In addition to the validation of FE model provided by Kim et al., Wang confirmed the validation of the optimum solution of MFs creating FCLs in various postures by comparing the active MFs of the model with previous studies. Their study implied that there exist numerous spinal MF combinations creating pure FCLs in the lumbar while maintaining the stable condition, which supports the hypothesis of FCLs as normal physiological loads in the spine in-vivo. In addition, it was revealed that the magnitudes of FCLs and the stability of the lumbar spine (i.e., deflection of the spinal column or trunk sway) might vary more sensitively to the force variation in short intrinsic muscles (SIMs) than to those in long spinal muscles, but it was not well quantified.

In this study, the effect of the maximum muscle force capacity (MFC) variation of long muscles and short intrinsic muscles (SIMs – interspinales, intertransversarii, and rotatores) on the feasibility of spinal MFs creating FCLs and stabilizing the spinal column are studied in five different postures, which are neutral standing, flexion 40°, extension 5°, left axial rotation 10°, and right lateral bending 30°.

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CHAPTER 3

METHOD

In order to investigate the effect of varying strength of lumbar spinal muscles on the stability of and the internal loads in the lumbar spine, computational analyses were performed using the same methods as used in a previous study based on 3-demensional (3-D) finite element (FE) and optimization models of a spinal system consisting of trunk, lumbar spine, sacrum-pelvis and spinal muscles [12, 24]. The geometrical data for an FE model were used geometrical input for an optimization analysis to predict spinal muscle forces (MFs) creating follower compressive loads (FCLs) in the lumbar spine in a given posture. Then, the optimum solutions of MFs were provided into a corresponding FE model as input forces producing the deformation (or deflection) of the spinal column. While no changes were made in FE models, inequality constraints of maximum muscle force capacity (MFC) in the optimization models were modified to simulate the muscles with various strengths in this study. Details of the computational models and the analysis method are presented in the following sections.

3.1 3-D Finite Element Analyses

In all FE models used in this study, bony structures (trunk, 6 lumbar vertebrae, and sacrum-pelvis) were considered rigid bodies connected by ligaments and intervertebral joints consisting of intervertebral disc and two facet joints. Rigid bodies were modeled using shell elements with extremely high elastic modulus. Facet joints

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were simulated using a non-linear compression-tension spring element to mimic their mechanical functions of resisting compression by bony contact as well as tension by capsular ligament. Intervertebral discs were modeled as a non-linear isotropic solid element. Mechanical roles of anterior and posterior longitudinal ligament (ALL, PLL) were incorporated in the disc material properties. Mechanical functions of the other remaining ligaments were represented as a single tension spring element located between the spinous processes since those can generate only tensile force by elongation.

A total of 244 spinal muscle fascicles (4 serratus posterior inferior, 14 latissimus dorsi, 12 external oblique, 12 internal oblique, 48 longissimus, 24 iliocostalis, 12 psoas major, 10 quadratus lumborum, 8 rectus abdominis, 6 spinalis thoracis, 40 multifidi, 12 interspinales, 20 intertransversarii, 22 rotatores) were modeled in the FE model. Spinal muscles can be grouped in many different ways, and one of the grouping criteria is the depth of the muscle from the skin as shown in Figure 3-1. Latissimus dorsi (LD), external oblique (EO), internal oblique (IO) and rectus abdominis (RA) are in a group of the superficial layer. The intermediate layer includes erector spinae (ES) and serratus posterior inferior (SPI). Rotatores, intertransversarii, interspinales, quadratus lumborum (QL), psoas major (PM) and multifidi are categorized as a deep muscle layer. Among the muscles in deep layer, rotatores, intertransversarii, and interspinales are classified as short intrinsic muscles (SIMs). In this study, all muscles other than SIMs are considered as a long muscle regardless of their length, orientation, or location for the purpose of parametric studies described in section 3.3. Each muscle fascicle was simulated by using one-dimensional discrete elements which all the application of a force of a constant magnitude between the insertion and origin points of each muscle so to allow the force

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direction adjustment along the muscle orientation that vary with the deformation of the spine.

The FE models of the spinal system in various postures were created by applying only a pure moment to the geometrical center (GC) of the T12 vertebral body in the FE model of the FE model of the spinal system in a neutral standing posture. For example, the spinal system in 40° flexion posture was created by applying a flexion moment was applied to the T12 of the spinal column in the neutral standing posture while holding the sacrum and pelvis fixed and applying no forces in all muscles. A total of 5 FE models for 5 different postures (neutral standing, 40° flexion, 5° extension, 10° axial rotation, and 30° lateral bending) were used in this study. The geometrical configurations of these FE models were used to formulate corresponding optimization problem formulation.

The optimization solutions of spinal MFs were used as force input data to the FE model of the spinal system in the corresponding posture in order to predict the deformation of the lumbar spine produced by the spinal MFs and the trunk weight of 350 N simulated as a vertical force applied at the CG of the trunk. The deformation of the lumbar spine was represented by the movement of the center of the gravity (GC) of the trunk, which was attached to the T12 vertebra rigidly. Such movement of the trunk GC was called “trunk sway” or “trunk displacement (TD)”. The magnitude of this TD vector was used as an indicator for the stability of the spine. As in the previous study, the spinal system was considered in a stable condition when the magnitude of TD ≤ 5 mm although there is no definite mechanical or physiological definition for the stability range of the spine deflection [12]. In fact, the trunk sway less than 5 mm was equivalent to the changes less than 1.5° in Cobb’s angle of the lumbar spine.

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All FE analyses were conducted using a commercial FE software, LS DYNA version 971.

3.2 Optimization Analyses of the Spinal System

Optimization problems to find spinal MFs creating follower compressive loads (FCL) in the lumbar spines in various postures were formulated and solved in the previous study [12]. The same optimization analysis methods were employed in this study. Briefly, the cost function was the summation of the magnitudes of JRFs and JRMs as used in the previous studies.

$$min(f) = \sum_{l=1}^{7} \| F_l^{jt} \| + \sum_{l=1}^{7} \| M_l^{jt} \| \quad (3.1)$$

, where F_l^{jt} and M_l^{jt} are the segmental JRF and JRM vectors in the *l*-th vertebra, respectively. In fact, the JRFs are equivalent to the FCLs whose directions are parallel to the lines connecting the GCs of the adjacent vertebral bodies as shown in Figure 3-1.

The constraints were as follows:

Force equilibrium equations:

$$\sum_{k=1}^{n} \vec{F}_{k,l}^{m} + \sum_{k=1}^{n'} \vec{F}_{k,l}^{ext} + \vec{F}_l^{jt} + \vec{F}_{l+1}^{jt} = 0 \quad (l = 1, \ldots, 6) \quad (3.2)$$

Moment equilibrium equations:

$$\sum_{k=1}^{n} \vec{r}_{k,l}^{m} \times \vec{F}_{k,l}^{m} + \sum_{k=1}^{n'} \vec{r}_{k,l}^{ext} \times \vec{F}_{k,l}^{ext} + \vec{\rho}_{k,l}^{jt} \times \vec{F}_l^{jt} + \vec{\rho}_{k,l+1}^{jt} \times \vec{F}_{l+1}^{jt} + \sum_{k=1}^{n''} \vec{M}_k^{jt} = 0 \quad (l = 1, \ldots, 6) \quad (3.3)$$

Equations to constrain the directions of JRFs parallel to the spinal curvature were:

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$$(\vec{r}_{l+1} - \vec{r}_l) \,//\, (\vec{d}_{l+1} - \vec{d}_l) \qquad (l = 1, \dots, 6);$$

(3.4)

$$\vec{\rho}_l = \vec{r}_l - \vec{d}_l \qquad (l = 1, \dots, 6)\;;$$

(3.5)

$$\| \vec{\eta}_l \| = \| \vec{\eta}_{l+1} \| = \| \vec{\eta} \| \qquad (l = 1, \dots, 6);\ \text{and}$$

(3.6)

$$\vec{\eta}_l \cdot (\vec{d}_{l+1} - \vec{d}_l) = 0.$$

(3.7)

Inequality constraints were as follows.

Range of FLP: $0 \leq \| \vec{\rho}_l \| \leq 15 \quad (l = 1, \dots, 6)$

(3.8)

Maximum muscle force capacity (MFC): $0 \leq \left\| \vec{F}_j^m \right\| \leq F_{max}^m \quad (j = 1, \dots, 244)$

(3.9)

Wang was able to find optimum solutions of spinal MFs that create the FCLs in the lumbar spine in each of 5 different postures with the minimum cost function value[12]. However, it was also found in FE analyses that the application of such optimum MFs results in the deflection of the lumbar spine large enough to make the trunk sway (or TD) greater than 10 mm. In contrast, the application of the spinal MFs creating the FCLs along the base FLP path ($\vec{\rho}_l = 0$ and $\vec{\eta}_l = 0$) produces the smallest deflection of the lumbar spine (TD < 5 mm) regardless of the postures of the lumbar spine. The optimization formulations

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used in this study included two more equality constraints ($\vec{\rho}_l = 0$ and $\vec{\eta}_l = 0$) to predict spinal MFs that can induce the smallest deflection of the lumbar spine.

For the formulation of an optimization problem, Geometrical data for a spinal system with the lumbar spine in a posture simulated in the corresponding FE model were used in this study. Such Geometrical data included: position and orientation of rigid bodies, points of insertion and origin of 244 muscle fascicles. Another input data was the trunk weight of 350 N applied at the trunk CG. Then, optimization problem was solved using commercial software, What's Best 11.0. 1.0 (Lindo Systems, Inc.) in order to determine the values of unknowns, such as 244 MFs, 6 segmental joint reaction forces (JRFs) whose direction follows the spinal curvature (i.e., follower compressive loads (FCLs)), 6 segmental joint reaction moment (JRM) vectors.

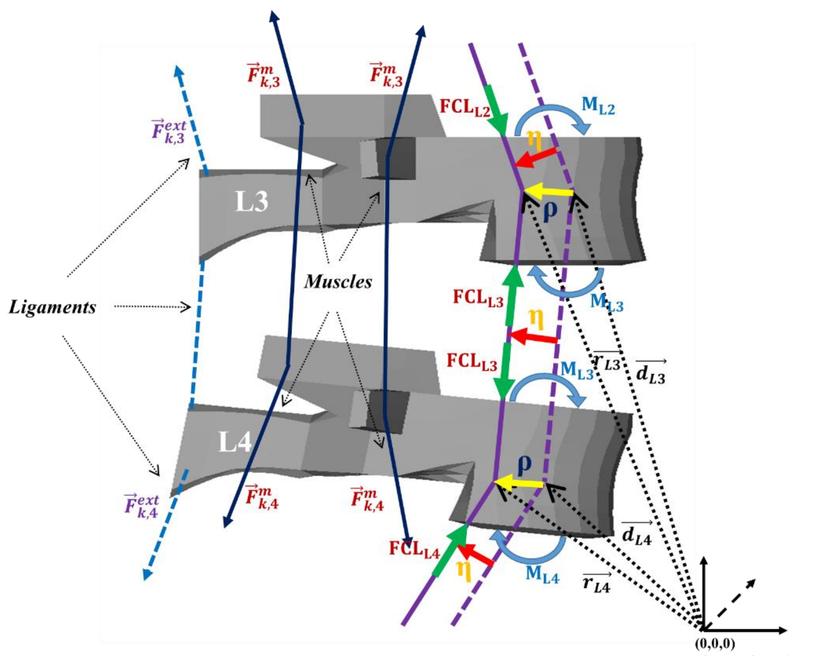

Figure 3-1 Schematic free body diagram for the optimization model. The JRFs directions are parallel to the lines connecting GCs of the vertebrae bodies to create the follower load constraint.

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3.3 Parametric studies

In order to investigate the effects of the strength of spinal muscles on the stability of the spinal system, FCL creating JRFs and TDs in the spinal systems with varying spinal muscle strengths were predicted and compared in this study. The maximum force that a muscle can produce used to be determined as a product of its physiological cross-sectional area (PCSA) and the maximum force capacity (MFC). Values of F_{max}^{m} in the optimization formulation were also determined in the same way. Because the physiological range of MFCs of spinal muscles was found to vary from 10 N/cm^2 to 90 N/cm^2 in previous studies [11], the spinal muscle strength variations were simulated changing the values of MFCs but with no changes in PCSAs in this study. Normal muscles were assumed to have median MFC (45 N/cm^2) while the increase in MFC indicates the muscle strengthening or *vice versa*.

For the parametric study, the spinal muscles were classified into short intrinsic muscles (SIMs) and long muscles. The MFCs of SIMs and long muscles were represented as MFC-S and MFC-L, respectively. MFC-S 100% (or MFC- L 100%) indicates the SIMs (or long muscles) with 45 N/cm^2. In this study, five different conditions of MFCs for both long muscles and SIMs (0%, 40%, 100%, 160%, and 200%) were evaluated in five different postures, which are neutral standing, flexion 40°, extension 5°, left axial rotation 10°, and right lateral bending 30° postures.

The magnitudes of joint reaction loads (JRFs and JRMs) and trunk displacements (TDs) predicted for 25 cases of MFC variation were compared in order to investigate the effect of the strength of spinal muscles on the stabilization of the lumbar spine in a given

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posture. A total of 125 cases (5 postures x 25 strengths of spinal muscles) were investigated computationally in this study.

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CHAPTER 4

RESULTS

Optimum solutions of spinal MFs creating the FCLs along the lumbar spinal curvature were not feasible when assuming no or small contributions of long muscles (i.e., MFC-L < 40%) regardless of the strength of SIMs (i.e., 0% ≤ MFC-S ≤ 200%). It was possible, however, to predict spinal MFs creating the FCLs whose direction follow the lines connecting the GCs of the adjacent vertebral bodies in all lumbar segments in cases of MFC-L ≥ 40% and MFC-S ≥ 0%. The trunk displacement (TD) resulting from the application of such spinal MFs and the trunk weight was predicted in terms of the displacement of the center of gravity (CG) of the trunk from FE models in order to estimate the stability of the lumbar spine in various 3-D postures. It was also possible to predict the changes in the spinal MFs, FCLs, and TDs due to variations in muscle strength (MFC changes) from 0 (no muscle force) to 0.9 MPa (maximum muscle force) in long muscles and/or SIMs. All results predicted for all 3-D postures of the lumbar spine simulated in this study are presented in the following sections.

4.1. Results from the Lumbar Spine in Neutral Posture

TDs predicted for the lumbar spine in neutral posture are shown in Figure 4-1. TDs were less than 1 mm regardless of the changes in strength of long muscles and SIMs (MFC-L and MFC-S in Figure 4-1, respectively) as long as both MFC-L and MFC-S are not zero. However, TDs were greater than 1 mm in cases of no SIM forces (MFC-S of 0%) and MFC-L 100% or greater while TD for the case of 0% MFC-S and 40% MFC-L

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was greater than 10 mm, which was considered as an unstable deformation of the lumbar spine. These results indicate that the neutral standing can be maintained stably with a minimal contribution of the spinal MFs as long as they create the FCLs in the spine.

Total joint reaction forces (JRFs) were determined as the summation of the magnitudes of FCLs in all segments (from T12-L1 to L5-Sa) as shown in Figure 4-2. When both the long muscles and SIMs have a good strength (MFC-L $\geq$ 40% and MFC-S $\geq$ 100%), the total JRFs were a little less than 4000 N with almost no changes with MFC variations. However, a substantial increase in total JRFs was predicted when MFC-S was 40% regardless of long muscle strength (MFC-L$\geq$ 40%). The total JRFs were predicted to exceed 6000 N with no contribution of SIMs (MFC-S = 0%) even in the cases of MFC-L $\geq$ 100%. In these cases, there was no joint reaction moments (JRMs) predicted in all lumbar segments. In contrast, in case of 40% MFC-L and 0% MFC-S, total JRF was about 6100 N but total JRM was 16000 Nmm in the direction of flexion, which may result in large TD > 10 mm as indicated by * in Figure 4-1. Figure 4-3 depicts the segmental JRFs in some selected cases of MFC variation.

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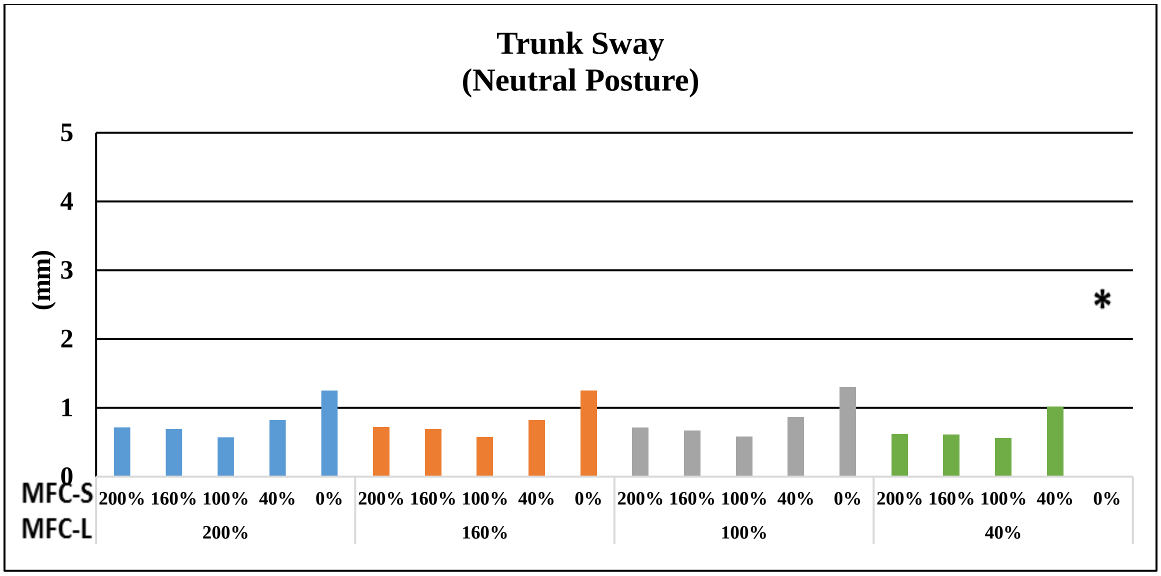

Figure 4-1 A prediction of trunk sways in MFC variation of SIMs and long muscles for neutral posture. Deflections which are less than 5mm were considered as stable conditions. (* Trunk sway > 10, buckling case)

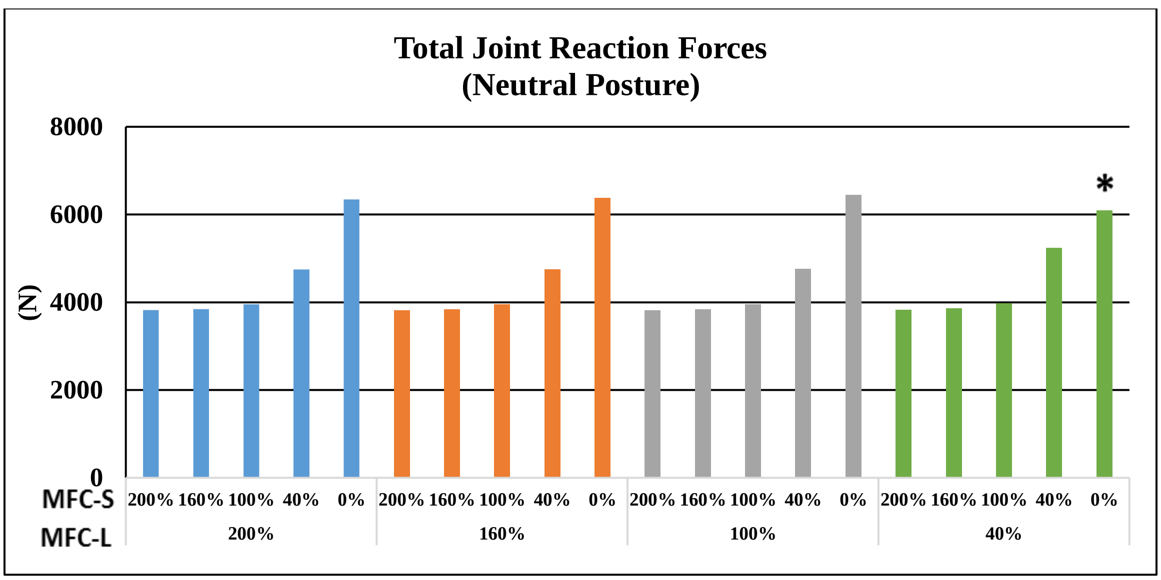

Figure 4-2 A prediction of total joint reaction forces (JRFs) in MFC variation of SIMs and long muscles for neutral posture. (* Trunk sway > 10, buckling case)

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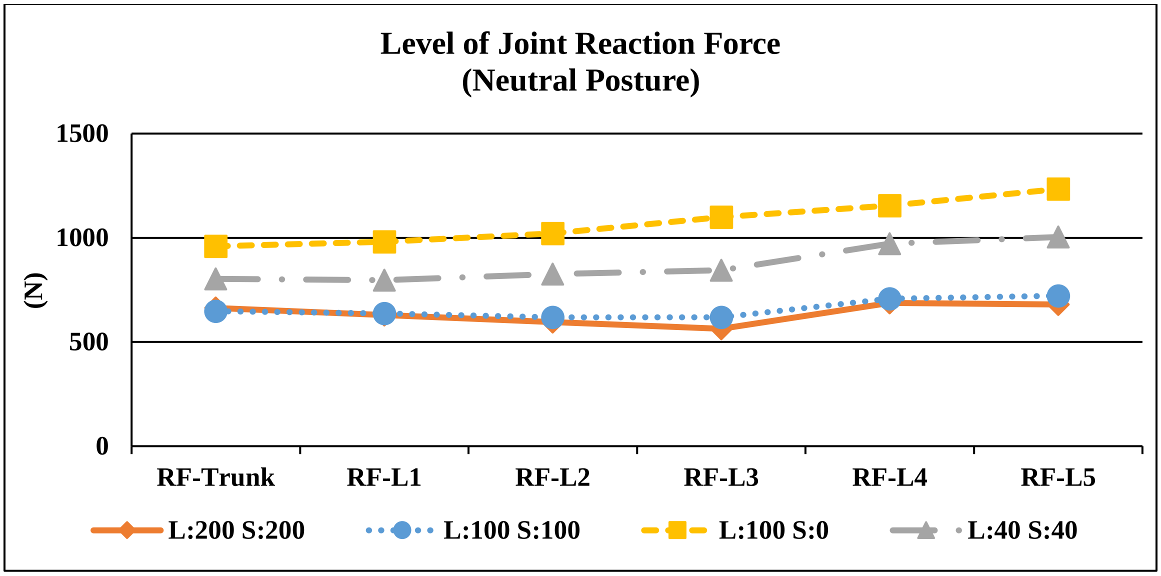

Figure 4-3 A variation of the level of joint reaction forces (JRFs) in some selected cases with MFC variation of SIMs and long muscles for neutral posture. (* Trunk sway > 10, buckling case)

Spinal MFs required for creating FCLs and maintaining the neutral posture in a stable condition are listed in Table 1. When compared with the case of 100% MFC-L and 100% MFC-S (i.e., assumed as average MFC), increases in both MFC-L and –S were found not to change the muscle recruitment pattern with small changes in MFs. However, the weakening of SIMs (MFC-S < 100%) were predicted to result in substantial changes in spinal MFs while no contribution of SIMs (MFC-S 0%) resulted in the recruitment of multifidus that were not necessary in the other cases of muscle strength condition.

These results of TD, total JRF and spinal MFs and their changes with the muscle strength variations indicate the followings. (1) The neutral standing posture can be

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maintained stably with a minimal contribution of the spinal MFs as long as they create the FCLs in the spine. (2) Contribution of long muscle and SIM forces are required for stabilizing the spine with smaller JRFs although contribution of SIM forces can reduce the total JRFs more effectively than that of long muscle forces.

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Table 4-1 Predicted non-zero muscle forces (N) in MFC variation of long muscles and SIMs for neutral posture.

		MFC-L	200%	200%	160%	100%	100%	40%	40%
		MFC-S	**200%**	**100%**	**200%**	**100%**	**0%**	**100%**	**40%**
Long Muscles	**Superficial**	**External Oblique**	**4**	**24**	**4**	**24**	**141**	**23**	**99**
		Internal Oblique	**71**	**58**	**71**	**58**	**90**	**59**	**63**
		Latissimus Dorsi	**252**	**262**	**252**	**262**	**393**	**207**	**223**
	Intermediate	**Longissimus**	**314**	**360**	**314**	**360**	**968**	**358**	**733**
	Deep	**Multifidus**	**0**	**0**	**0**	**0**	**140**	**0**	**0**
		SUM	**641**	**704**	**641**	**704**	**1732**	**647**	**1118**
SIMs		**Interspinales**	**197**	**242**	**197**	**242**	**0**	**234**	**136**
		Intertransversarii	**149**	**180**	**149**	**180**	**0**	**180**	**180**
		Rotatores	**505**	**287**	**505**	**287**	**0**	**358**	**112**
		SUM	**850**	**709**	**851**	**709**	**0**	**773**	**428**
Total MFs			**1491**	**1413**	**1492**	**1413**	**1732**	**1420**	**1546**

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4.2 Results from the Lumbar Spine in 40°Flexion Posture

Figure 4-4 depicted TDs of the lumbar spine predicted from FE analyses of the spine in 40°flexion. TDs were less than 1mm when both long muscles and SIMs were strong enough (MFC-L and -S ≥ 100%). When MFC-L was greater than 100%, TDs were also less than 1mm regardless of the MFC of SIMs. In contrast, with the average capacity of the long muscles (MFC-L = 100%), the weakening of SIMs (MFC-S < 100%) increased TDs substantially to 2.25mm although still small enough to keep the flexed posture stably. TDs for the cases of 40% MFC-L were predicted to increase inversely with the weakening of SIMs (decreasing MFC-S) from about 3.5 mm (200% MFC-S) to 8 mm (40% MFC-S). When MFC-L and –S were 40% and 0%, respectively, the buckling of the lumbar spine in the direction of flexion was predicted as in the analyses of the neutral posture. These results implies that, unlike the neutral posture, the maximum capacity of long muscles is more crucial for maintaining the flexed posture compared to that of SIMs although both long muscles and SIMs were needed for creating FCLs and stabilizing the spine.

Total JRFs were shown in Figure 4-5. The overall variation of total JRFs as a function of muscle strength (MFC-L and –S) was similar to that of TDs. In the cases of stronger long muscles (MFC-L ≥ 160%), total JRFs were about 4000 N with small increases when MFC-S decreased from 200% to 0%. When MFC-L was 100%, total JRFs were predicted to vary from 4000 N to 6000 N with the weakening of SIM strength (200% to 0% MFC-S). When MFC-L was 40%, the predicted total JRF was almost 7000 N in case of 200% MFC-S and increased with the weakening of SIM strength up to 9500

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N (0% MFC-S). The buckling of the lumbar spine predicted in the case of 40% MFC-L and 0% MFC-S indicates that the critical load of lumbar spine in 40° flexion may be 9500 N. The JRFs in all motion segments from some stable cases were shown in Figure 4-6. When both long muscles and SIMs were in good condition (MFC-L and MFC-S > 100%), the segmental JRFs were similar in all lumbar levels. However, the weakening of SIM strength was predicted to increase the segmental JRF in the lower lumbar levels compared to the upper levels when MFC-L was low.

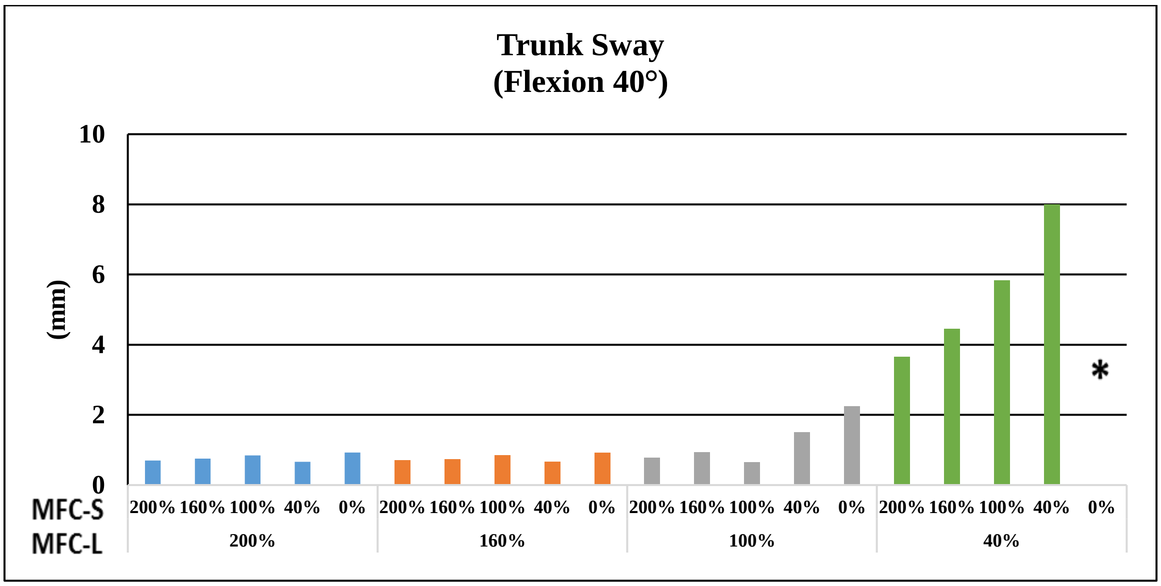

Figure 4-4 A prediction of trunk sways in SIMs and long muscles MFC variation for flexion 40°. Deflections which are less than 5mm were considered as stable conditions. (* Trunk sway > 10, buckling case)

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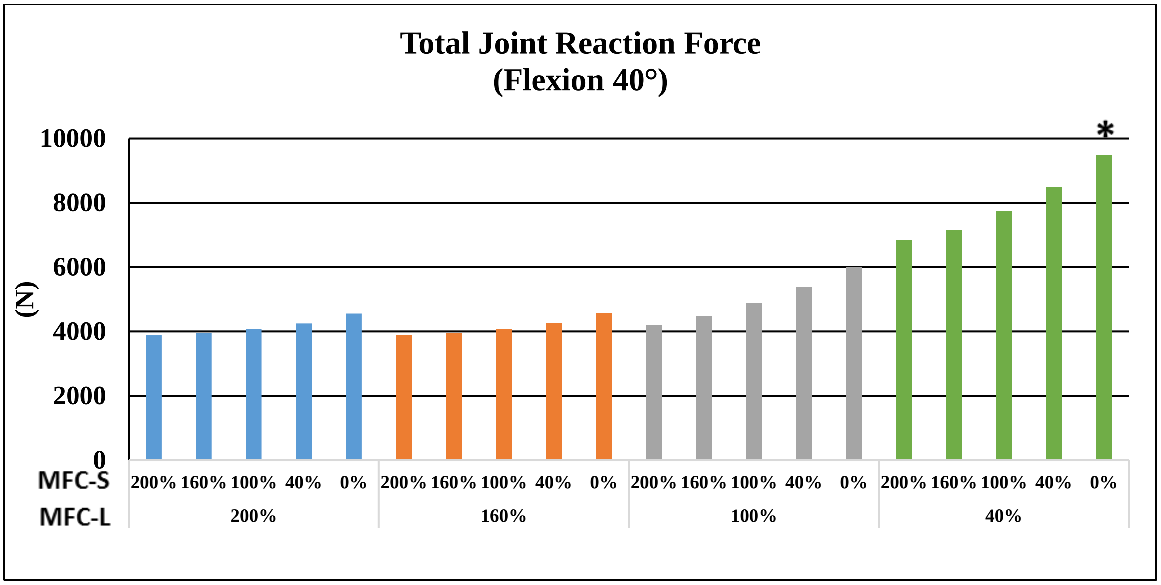

Figure 4-5 A prediction of total joint reaction forces (JRFs) in MFC variation of SIMs and long muscles for flexion 40°. (* Trunk sway > 10, buckling case)

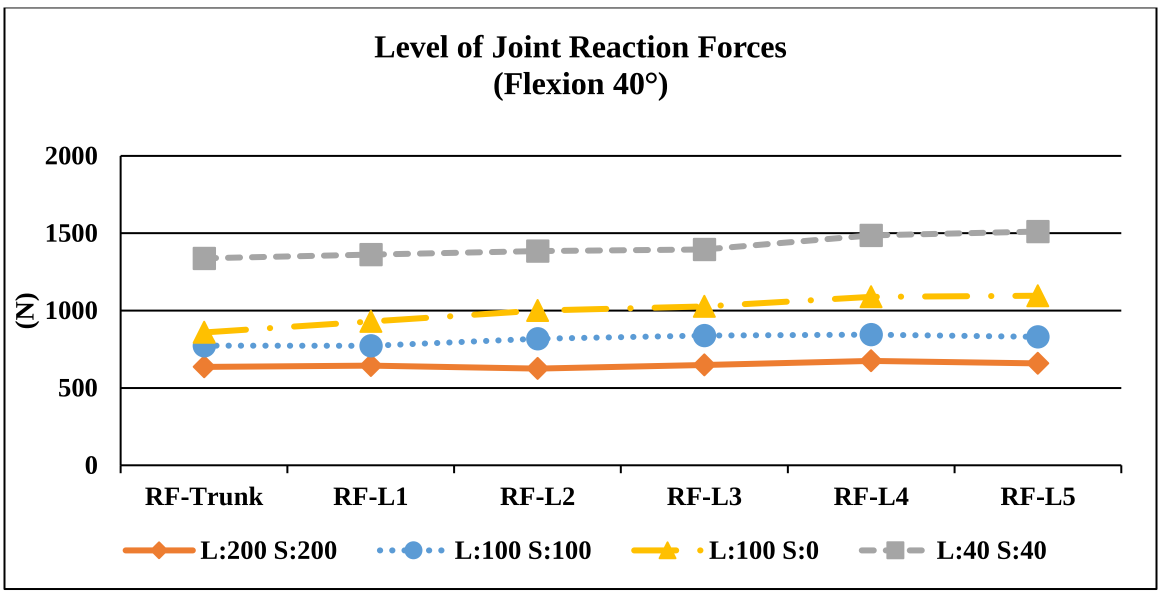

Figure 4-6 A variation of the level of joint reaction forces (JRFs) in some selected cases with (rewrite this legend) MFC variation of SIMs and long muscles for flexion 40°. (* Trunk sway > 10, buckling case)

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An active muscle types with non-zero MF values were shown in Table 4-2. When compared with the case of an average muscle capacity in both long and short muscles (100% MFC-L and S), it was predicted that the strengthening of SIMs increased the MFs in SIMs but decreased MFs in long muscles although such changes in MFs were not significant. On the other hand, MFs in long muscles, such as in longissimus and LD, were increased significantly by weakening of SIMs and/or long muscles. , It is also interesting to note that Serratus Posterior Inferior muscles were newly recruited in cases of MFC-S $\leq$ 40%, while not predicted in the other cases of MFC conditions.

The results of TD, total JRFs, and MFs predicted in variation of MFC-L and S indicate the followings. (1) Flexion 40° postures can be maintained stably by MFs creating FCLs in the lumbar spine. (2) While the MFs of both long muscles and SIMs are required to keep the posture, long muscles play more crucial role to reduce not only the trunk deflection but also the total JRFs.

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Table 4-2 Predicted non-zero muscle forces (N) in MFC variation of long muscles and SIMs for flexion 40°.

.		MFC-L	200%	200%	160%	100%	100%	40%	40%
		MFC-S	**200%**	**100%**	**0%**	**100%**	**0%**	**100%**	**40%**
Long Muscles	**Superficial**	**External Oblique**	**214**	**235**	**278**	**176**	**176**	**70**	**70**
		Internal Oblique	**10**	**8**	**15**	**0**	**0**	**27**	**7**
		Rectus Abdominis	**0**	**0**	**0**	**31**	**61**	**151**	**197**
	Intermediate	**Iliocostalis**	**0**	**0**	**0**	**0**	**1**	**85**	**141**
		Longissimus	**81**	**91**	**108**	**237**	**402**	**614**	**672**
		Spinalis Thoracis	**101**	**170**	**85**	**157**	**114**	**191**	**209**
		Serratus Posterior Inferior	**0**	**0**	**85**	**0**	**97**	**0**	**31**
	Deep	**Multifidus**	**120**	**129**	**183**	**211**	**183**	**222**	**243**
		Psoas Major	**13**	**77**	**212**	**59**	**133**	**25**	**51**
		Quadratus Lumborum	**94**	**98**	**106**	**32**	**74**	**2**	**0**
		SUM	**633**	**808**	**1072**	**903**	**1241**	**1387**	**1621**
SIMs		**Interspinales**	**121**	**142**	**0**	**65**	**0**	**24**	**69**
		Intertransversarii	**296**	**123**	**0**	**149**	**0**	**249**	**108**
		Rotatores	**71**	**46**	**0**	**95**	**0**	**84**	**69**
		SUM	**488**	**311**	**0**	**309**	**0**	**357**	**246**
Total MFs			**1121**	**1119**	**1072**	**1212**	**1241**	**1744**	**1867**

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4.3 Results from the Lumbar Spine in 5° Extension Posture

The trunk displacements (TDs) predicted from the cases of various MFC-S and MFC-L were shown in Figure 4-7. As long as MFC-L and S were 100% or greater, TDs were less than 2 mm. When long muscles were strong enough (i.e., MFC-L ≥ 100%), TDs for the cases of 40% MFC-S were substantially increased but still less than 5mm. When the long muscles capacity were 40%, TDs of the cases MFC-S ≥ 100% were less than 5 mm although TD with 100% MFC-S was increased by 170%. In contrast, when no contribution of SIMs (i.e., MFC-S 0%) was simulated, the buckling of the lumbar spine was predicted even in the cases with MFC-L≥100% (* in Figure 4-7). When the strength of long muscles is small (MFC-L 40%), the lumbar spine was predicted to buckle in the case of MFC-S 40%), and no optimum solution for FCL creating MFs was feasible in the case of MFC-S 0%.

Figure 4-8 shows that total JRF was predicted to increase with decreasing strength of spinal muscles as predicted in the neutral standing and flexed postures. In 5° extension posture, magnitudes of JRFs were bigger than those in the neutral standing and flexed postures. The increase in total JRF was large in cases of weak strength of SIMs (MFC-S ≤40%) while it was moderate in cases of strong muscle strength (MFC-L ≥ 40% and MFC-S ≥ 100%). JRFs in each level of selected cases in variation of MFCs are shown in Figure 4-9.

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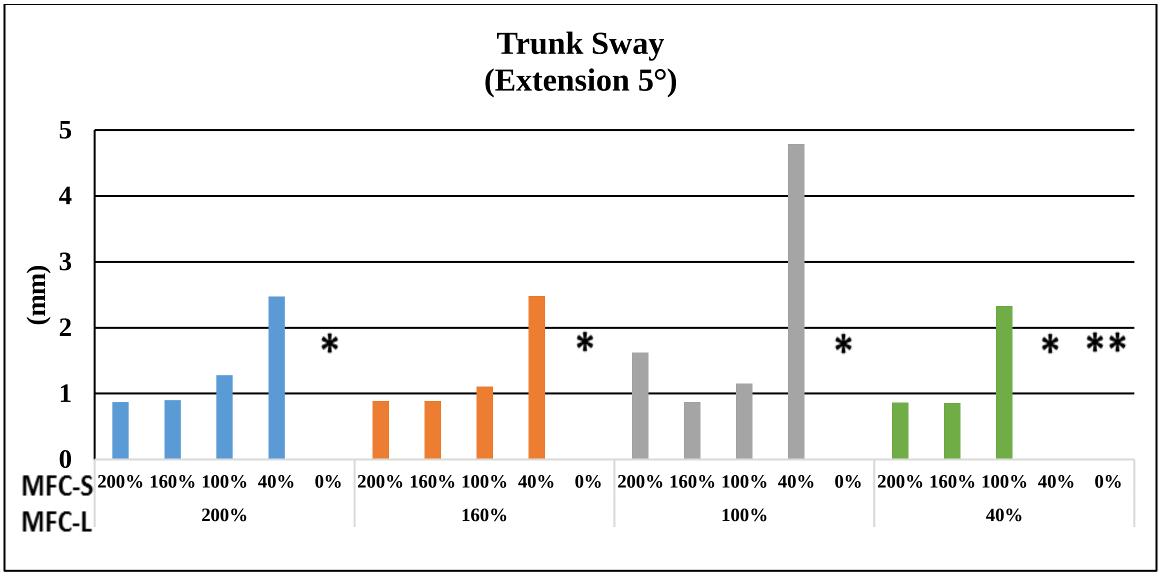

Figure 4-7 A prediction of trunk sways in SIMs and long muscles MFC variation for extension 5°. Deflections which are less than 5mm were considered as stable conditions. (* Trunk sway > 10, buckling case), (** No optimum solution case)

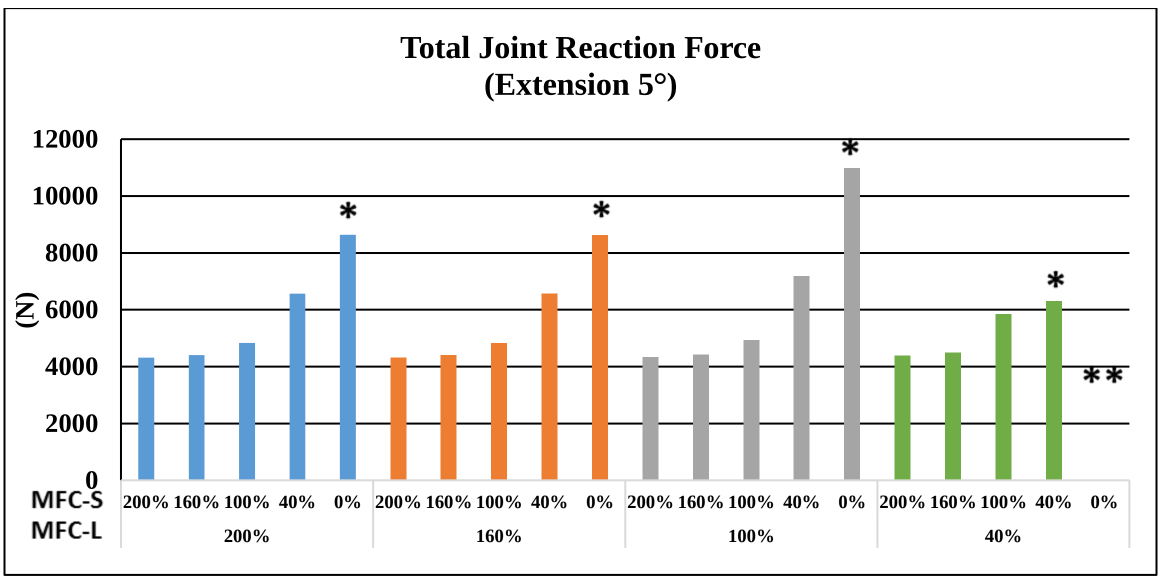

Figure 4-8 A prediction of total joint reaction forces (JRFs) in MFC variation of SIMs and long muscles for extension 5°. (* Trunk sway > 10, buckling case), (** No optimum solution case)

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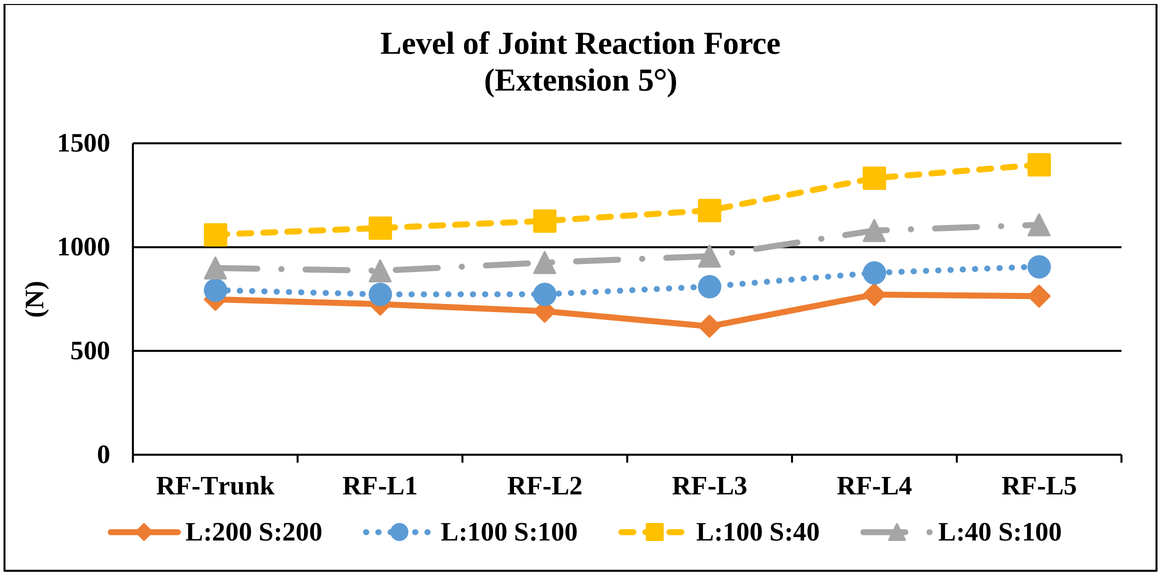

Figure 4-9 A variation of the level of joint reaction forces (JRFs) in some selected cases with MFC variation of SIMs and long muscles for extension 5°. (* Trunk sway > 10, buckling case)

Non-zero spinal MFs required for maintaining the extension 5° posture in a stable manner in cases of variable muscle strengths were presented in Table 4-3. Compared to the case in average muscle capacity (100% MFC-L and S), no significant changes in MFs was observed by the strengthening of long muscles as well as SIMs. In contrast, the reduction of MFC-L resulted in the substantial changes in MFs, and iliocostalis and spinalis thoracis were predicted to be newly recruited. The weakening of SIMs (MFC-S = 40%) resulted in the significant increase of MFs, and mainly, considerable increase of MFs were observed in longissimus.

These results indicate the followings. (1) The extension posture can be maintained in a stable manner with the contribution of spinal MFs by creating FCLs in the lumbar spine. (2) Contribution of long muscle is important to reduce the load on the spine, but the contribution of SIMs is crucial for the stability of the extended spine. (3)

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The stable maintenance of the extended posture is more challenging than that of the neutral standing posture and the flexed posture.

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Table 4-3 Predicted non-zero muscle forces (N) in MFC variation of long muscles and SIMs for extension 5°.

		MFC-L	200%	200%	160%	100%	100%	100%	40%
		MFC-S	**200%**	**100%**	**40%**	**200%**	**100%**	**40%**	**100%**
Long Muscles	**Superficial**	**External Oblique**	**54**	**81**	**165**	**61**	**99**	**218**	**154**
		Internal Oblique	**69**	**64**	**82**	**64**	**57**	**75**	**64**
		Latissimus Dorsi	**333**	**320**	**562**	**290**	**353**	**375**	**224**
	Intermediate	**Iliocostalis**	**0**	**0**	**0**	**0**	**0**	**0**	**24**
		Longissimus	**382**	**471**	**870**	**399**	**523**	**1176**	**739**
		Spinalis Thoracis	**0**	**0**	**0**	**0**	**0**	**0**	**69**
	Deep	**Multifidus**	**0**	**33**	**121**	**0**	**0**	**35**	**1**
		SUM	**838**	**969**	**1800**	**814**	**1032**	**1879**	**1275**
SIMs		**Interspinales**	**375**	**334**	**144**	**384**	**338**	**144**	**336**
		Intertransversarii	**181**	**372**	**269**	**217**	**384**	**251**	**447**
		Rotatores	**570**	**448**	**144**	**590**	**370**	**113**	**233**
		SUM	**1126**	**1154**	**557**	**1191**	**1092**	**508**	**1016**
Total MFs			**1964**	**2123**	**2357**	**2005**	**2124**	**2387**	**2291**

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4.4 Results from the Lumbar Spine in 10° Left Axial Rotation

TDs predicted from the cases with various MFC-L and S were shown in Figure 4-10. In cases of MFC-S ≥ 40%, TDs were less than 2 mm in cases of MFC-L ≥ 100%), whereas TDs were greater than 2 mm in cases of no SIM contributions (MFC-S = 0%). In particular, the TD for the case of MFC-L = 100% and MFC-S = 0% was greater than 6 mm. The TD for the case of weak strength of both long muscles and SIMs (MFC-L = 40% and MFC-S = 40%) was greater than 10 mm. In the case of weak strength of long muscles (MFC-L = 40%) and no contribution of SIMs (MFC-S = 0%), anterior buckling of the lumbar spine was predicted from the FE analysis although the optimum solutions were feasible. These results indicate the crucial role of SIMs for the stable maintenance of the lumbar spine in the posture of axial rotation 10°.

Figure 4-11 showed total JRFs in the lumbar spine in axially rotated by 10°. In the cases of axially rotated posture, the optimization analyses predicted non-zero joint reaction moments (JRMs) in cases of no contribution of SIMs (Table 4-4). Regardless of MFC-L, total JRFs in cases of MFC-S ≥ 100% were less than 4650N. In the cases of MFC-S 40% and MFC-L ≥ 100%, total JRFs were increased up to 6000N while TDs were still less than 1mm. In these cases of small TDs, no JRMs were predicted in all lumbar levels. In contrast, in cases of MFC-L ≥ 100% and MFC-S = 0% (i.e., no contribution of SIMs), total JRFs were greater than 8000N and JRMs were predicted in a few lumbar levels as shown in Table 4-4. In the cases of weak muscles strength (MFC-L ≤ 40% and MFC-S ≤ 40%), the values of total JRF were about 6000 N (Figure 4-11) but

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the greater JRMs were predicted in more spinal levels (Table 4-4). The selected cases of JRFs in each level were shown in figure 4-13.

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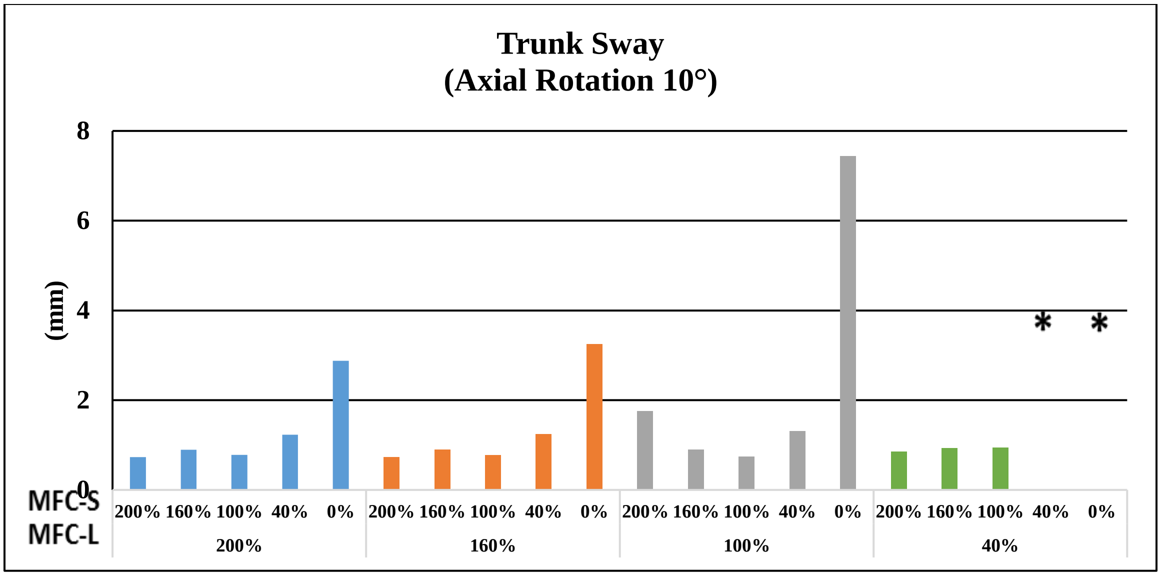

Figure 4-10 A prediction of trunk sways in SIMs and long muscles MFC variation for left axial rotation 10°. Deflections which are less than 5mm were considered as stable conditions. (* Trunk sway > 10, buckling case)

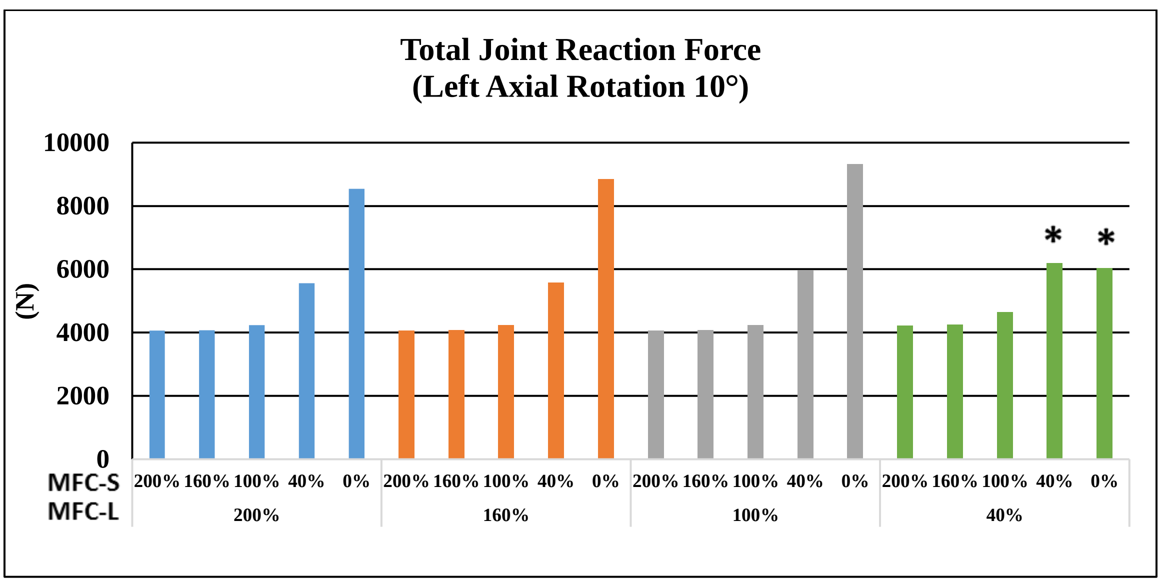

Figure 4-11 A prediction of total joint reaction forces (JRFs) in MFC variation of SIMs and long muscles for left axial rotation 10°. (* Trunk sway > 10, buckling case)

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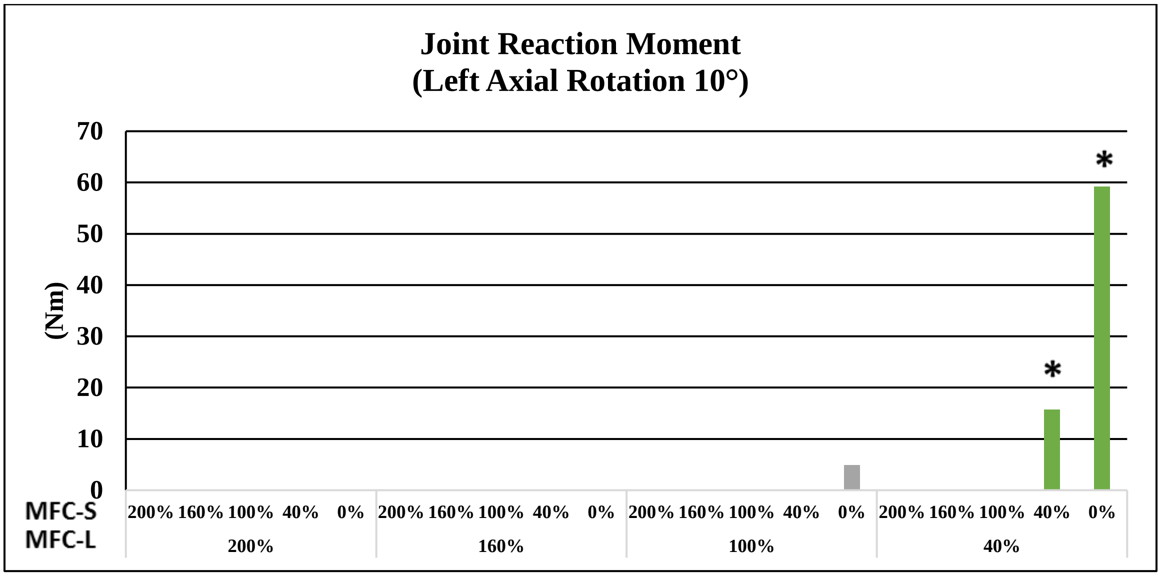

Figure 4-12 A prediction of total joint reaction moments (JRMs) in MFC variation of SIMs and long muscles for left axial rotation 10°. (* Trunk sway > 10, buckling case)

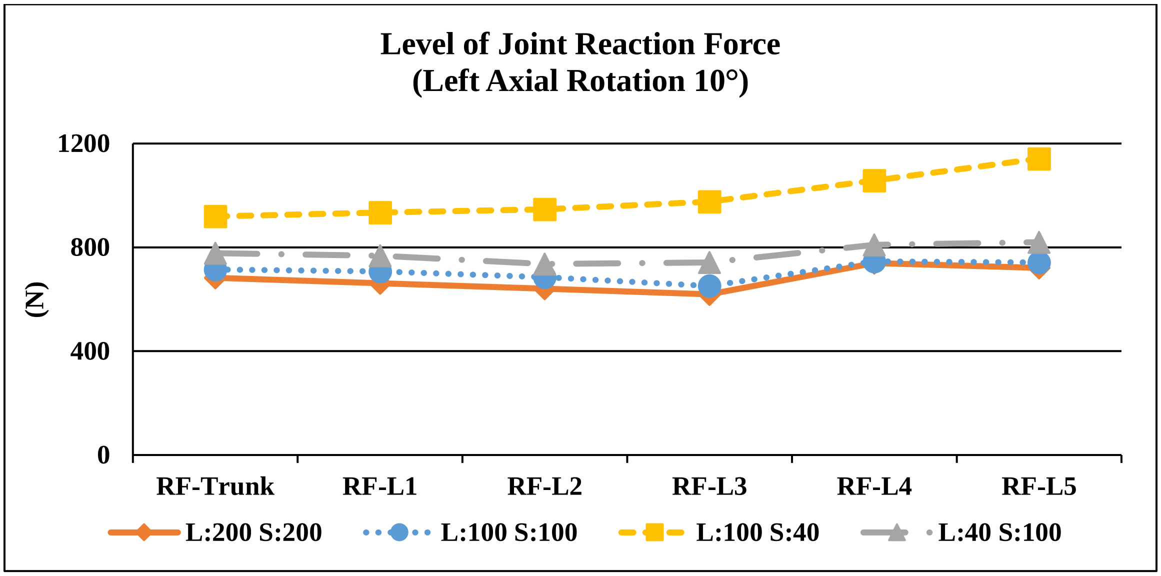

Figure 4-13 A variation of the level of joint reaction forces (JRFs) in some selected cases with MFC variation of SIMs and long muscles for left axial rotation 10°. (* Trunk sway > 10, buckling case)

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Non-zero MFs for maintaining the axial rotation posture and creating FCLs on lumbar spine are presented in Table 4-5. When compared to the case of average MFC (100% MFC-L and S), no significant change of MFs by strengthening of long muscles and/or SIMs was predicted. Likewise, in the case of weakening of long muscles (MFC-L = 40%), rectus abdominis was newly recruited. However, when MFC-S was decreased (MFC-S $\leq$ 40%), increases in MF values of the long muscles were predicted significantly. Psoas major and quadratus lumborum were predicted to be recruited, which were not predicted in the other cases.

The results of TDs, total JRFs and JRMs, and non-zero MFs predicted from the lumbar spine in 10° axial rotation posture indicate the followings. (1) The lumbar spine in 10° axial rotation posture can be maintained in a stable manner with FCL creating spinal MFs. (2) Contribution of both long and short muscles were required for maintaining the posture, but the effect of the reduction of MFC-S was more significant than that of MFC-L.

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Table 4-4. A Predicted non-zero JRMs (Nmm) in MFC variation of long muscles and SIMs for left axial rotation 10°.

MFC-L	**200%**	**160%**	**100%**	**40%**
MFC-S	**0%**	**0%**	**0%**	**40%**
Tr-Mz	**62**	**54**	**565**	**782**
L1-Mz	**0**	**0**	**470**	**663**
L2-My	**0**	**0**	**1354**	**6291**
L3-Mx	**0**	**0**	**0**	**819**
L3-My	**0**	**0**	**0**	**2623**
L3-Mz	**0**	**0**	**0**	**200**
L4-Mx	**0**	**0**	**572**	**0**
L4-My	**0**	**0**	**1985**	**4359**
Total JRM	**62**	**54**	**4946**	**15737**

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Table 4-5. Predicted non-zero muscle forces (N) in MFC variation of long muscles and SIMs for left axial rotation 10°.

		MFC-L	**200%**	**200%**	**160%**	**160%**	**100%**	**100%**	**40%**
		MFC-S	**200%**	**100%**	**40%**	**0%**	**100%**	**0%**	**100%**
Long Muscles	**Superficial**	**External Oblique**	**44**	**50**	**115**	**233**	**49**	**278**	**63**
		Internal Oblique	**46**	**47**	**66**	**146**	**48**	**156**	**39**
		Latissimus Dorsi	**306**	**281**	**459**	**634**	**274**	**423**	**154**
		Rectus Abdominis	**0**	**0**	**0**	**0**	**0**	**4**	**21**
	Intermediate	**Longissimus**	**278**	**296**	**529**	**1304**	**291**	**1630**	**377**
	Deep	**Multifidus**	**72**	**74**	**174**	**276**	**76**	**125**	**78**
		Psoas Major	**0**	**0**	**0**	**34**	**0**	**0**	**0**
		Quadratus Lumborum	**0**	**0**	**0**	**34**	**0**	**56**	**0**
		SUM	**746**	**748**	**1343**	**2661**	**738**	**2672**	**732**
SIMs		**Interspinales**	**240**	**201**	**144**	**0**	**201**	**0**	**320**
		Intertransversarii	**241**	**254**	**213**	**0**	**257**	**0**	**404**
		Rotatores	**434**	**376**	**149**	**0**	**403**	**0**	**373**
		SUM	**915**	**831**	**506**	**0**	**861**	**0**	**1097**
Total MFs			**1661**	**1579**	**1849**	**2661**	**1599**	**2672**	**1829**

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4.5 Results from the Lumbar Spine in 30° Right Lateral Bending

Trunk displacements (TDs) of lumbar spine in 30° right lateral bending are shown in Figure 4-14. TDs were predicted to less than 1mm in cases of MFC-L ≥ 160% regardless of MFC-S. TDs in case of average MFC-L (MFC-L = 100 %) were predicted to vary with MFC-S changes from about 1 mm to 2.06 mm. In all cases of weak strength of long muscles (MFC-L = 40%), however, FE analyses predicted the buckling of the spine into the ipsilateral direction even with good strength of SIMs (MFC-S ≥ 100%). No optimum solutions for FCL creating MFs were feasible in cases of weak strength of both long muscles (MFC-L = 40%) and SIMs (MFC-S ≤ 40%).

Figure 4-15 and Table 4-6 show total joint reaction forces (JRFs) and joint reaction moments (JRMs), respectively. Total JRFs in cases of MFC-L ≥ 100% were less than 5500N regardless of MFC-S. Moreover, in cases of greater MFC-L, lower total JRFs were observed. Likewise, total JRMs were increased by the reduction of MFC-L and/or MFC-S. No JRMs were predicted in cases of 160% MFC-L with MFC-S ≥ 160% and 200% MFC-L with MFC-S ≥ 100%. When MFC-L was 100% or greater, total JRMs were predicted to less than 17.2Nm. In cases of MFC-L = 40%, total JRFs were less than 5500N, but total JRMs were predicted up to 52 Nm when MFC-S was 100%. JRF variations in each lumbar level in some selected cases are shown in Figure 4-16.

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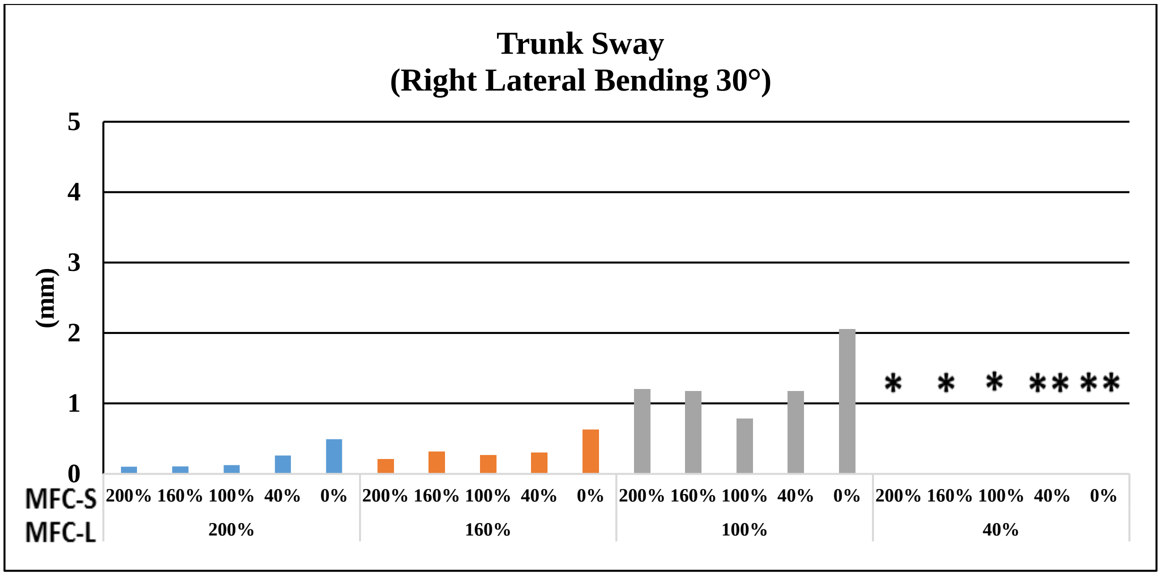

Figure 4-14 A prediction of trunk sways in SIMs and long muscles MFC variation for right lateral bending 30°. Deflections which are less than 5mm were considered as stable conditions. (* Trunk sway > 10, buckling case), (** No optimum solution case)

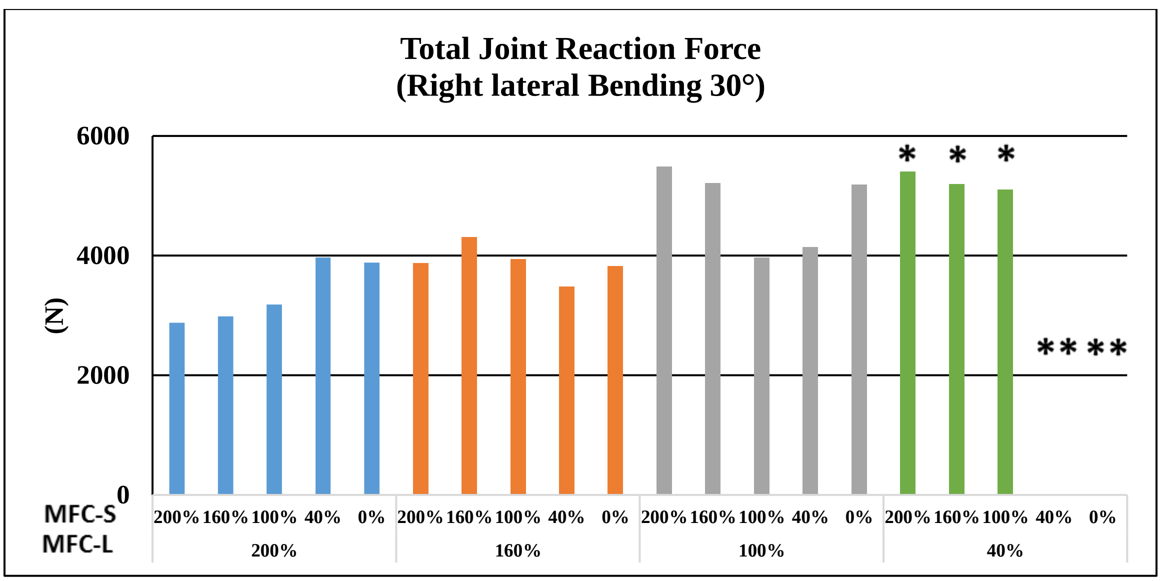

Figure 4-15 A prediction of total joint reaction forces (JRFs) in MFC variation of SIMs and long muscles for right lateral bending 30°. (* Trunk sway > 10, buckling case), (** No optimum solution case)

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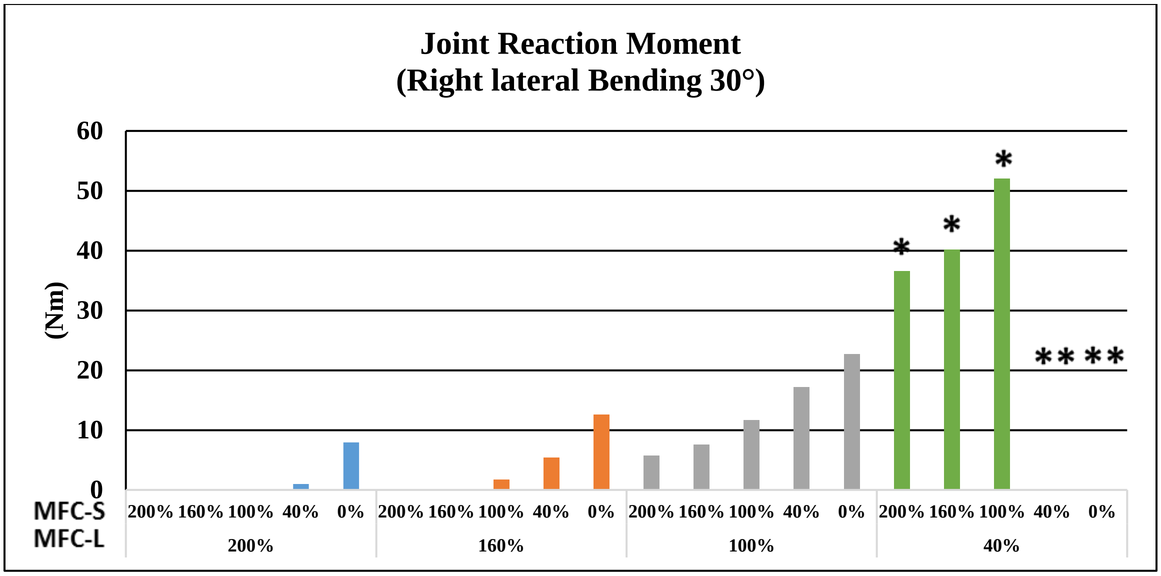

Figure 4-16 A prediction of total joint reaction moments (JRMs) in MFC variation of SIMs and long muscles for right lateral bending 30°. (* Trunk sway > 10, buckling case), (** No optimum solution case)

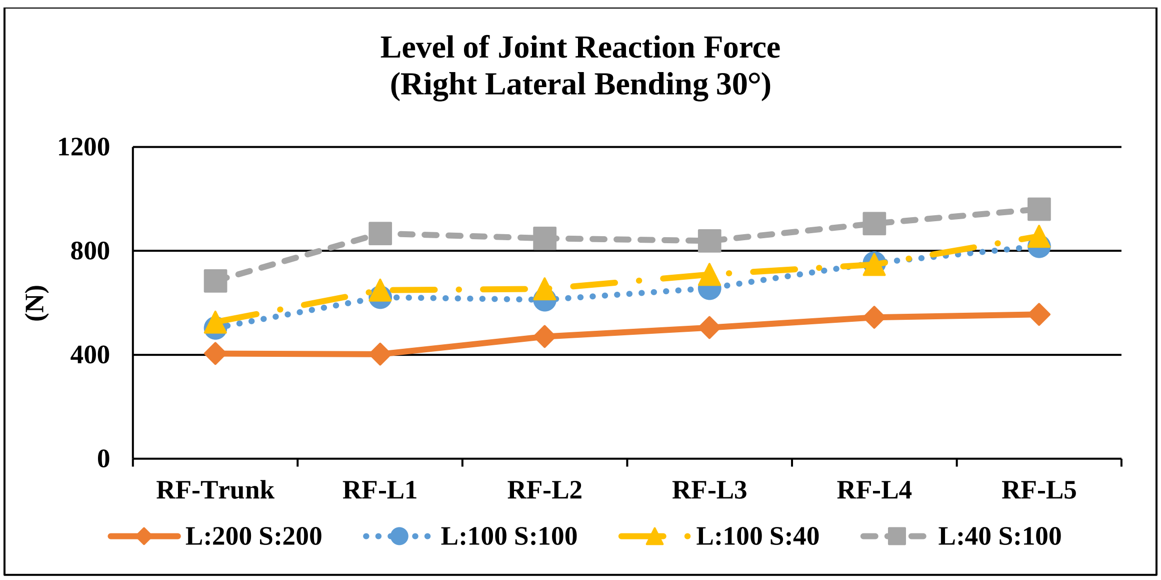

Figure 4-17 A variation of the level of joint reaction forces (JRFs) in some selected cases with MFC variation of SIMs and long muscles for right lateral bending 30°.

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Non-zero muscle forces for maintaining the spine in lateral bending posture are introduced in table 4-7. The magnitudes of the forces in long muscles were predicted to change minimally with the variation of the strength of both long muscles and SIMs. When compared to the maximum strength of long muscles (MFC-L = 200%), the weakening of the long muscles (MFC-L $\leq$ 160%) induced the increase of MFs of SIMs. In contrast, in cases of the strengthening of SIMs (MFC-S $\geq$ 160%), the significant MF increase of SIMs was predicted.

The results predicted from the lumbar spine in 30° right lateral bending indicate the followings. (1) It is possible to maintaining the lateral bending 30° posture in a stable manner by the spinal MFs creating the FCLs in the lumbar spine. (2) While forces in both long muscles and SIMs are required to keep the laterally bent posture stably, the contribution of long muscles may be more crucial to keep the posture.

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Table 4.6 A Predicted non-zero JRMs (Nmm) in MFC variation of long muscles and SIMs for right lateral bending 30°.

MFC-L	200%	200%	160%	160%	160%	100%	100%	100%	100%	100%	40%	40%	40%
MFC-S	40%	0%	100%	40%	0%	200%	160%	100%	40%	0%	200%	160%	100%
Tr-My	609	4123	1754	4418	4226	5241	5955	6212	6113	6691	9319	8670	9056
L1-My	0	0	0	0	0	0	0	0	0	0	752	469	0
L2-My	0	0	0	0	370	0	0	0	0	171	0	0	0
L2-Mz	277	1269	0	227	2360	0	430	1910	2682	3293	3144	3458	4521
L3-Mx	0	0	0	0	0	0	0	0	0	0	0	0	2019
L3-Mz	0	898	0	0	2168	524	771	1959	3953	4065	6837	6731	7117
L4-Mx	0	1182	0	0	1445	0	0	0	0	3392	982	3696	7030
L4-Mz	116	449	0	779	2023	0	374	1535	4466	5106	8309	7210	7726
L5-Mx	0	0	0	0	0	0	0	0	0	0	606	4663	9323
L5-Mz	0	0	0	0	0	0	0	0	0	0	6582	5281	5214
Total JRM	1003	7921	1754	5423	12592	5766	7530	11616	17214	22718	36531	40177	52007

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Table 4-7. Predicted non-zero muscle forces (N) in MFC variation of long muscles and SIMs for right lateral bending 30°.

		MFC-L	200%	200%	160%	100%	100%	100%	40%
		MFC-S	200%	100%	0%	200%	100%	0%	100%
Long Muscles	Superficial	External Oblique	170	191	218	247	223	280	178
		Internal Oblique	235	256	246	205	207	188	127
		Latissimus Dorsi	278	346	321	208	214	229	123
	Intermediate	Iliocostalis	5	16	87	23	20	204	42
		Longissimus	269	320	510	504	338	689	388
		Serratus Posterior Inferior	51	38	0	99	20	0	119
	Deep	Multifidus	37	70	103	210	144	202	224
		Psoas Major	81	115	144	223	103	181	50
		Quadratus Lumborum	82	71	27	134	79	5	0
		SUM	1208	1423	1656	1853	1348	1978	1251
SIMs		Interspinales	10	36	0	374	231	0	265
		Intertransversarii	362	189	0	763	388	0	589
		Rotatores	290	199	0	286	284	0	269
		SUM	662	424	0	1423	903	0	1123
Total MFs			1870	1847	1656	3276	2251	1978	2374

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CHAPTER 5

DISCUSSION

It is generally known that the current understandings of the basis of LBP problems are limited and insufficient but their roots are closely related to the mechanical insufficiency of the spinal column. A human ligamentous spine is a complex and inherently unstable column because it consists of flexible segments. However, it is generally agreed that such an unstable spinal column serves its biomechanical functions in a stable manner in-vivo with the support of spinal muscle forces resulting from an adequate motor unit control. While the motor unit control mechanism remains largely unknown, one possible mechanism suggested in recent studies is that the spinal MFs creating the follower compressive loads (FCLs) in the spinal column can stabilize the spinal column. Patwardhan et al. showed that the ligamentous lumbar spine supports the FCL up to 1200 N without buckling while maintaining its flexibility reasonably well [8, 38]. They also demonstrated the feasibility of the FCLs created by spinal MFs using a simple analytical model of a flexible column [39] However, only 10 muscle forces were simulated in the model, which was too much simplification to test if it is possible for spinal muscles create the FCLs in the lumbar spine. Kim and Kim published the results of a computational study showing that it is not feasible to create FCLs in the lumbar spine despite of considering more than 100 spinal MFs in their model [40].

On the other hand, there have been multiple computational studies demonstrating the feasibility for spinal MFs creating FCLs in the lumbar spine using FE and optimization analyses of the lumbar spine under the influence of 161 pairs of spinal muscles (including 27 pairs of SIMs) [12, 24, 37]. Most recently, Wang was able to

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confirm the feasibility of such FCL creating spinal MFs in the lumbar spine in various postures of flexion, extension, lateral bending and axial rotation [12]. Wang also showed that the deflection of the spinal column was minimal under the action of spinal MFs that produce the FCLs along the line connecting the GCs of the adjacent vertebral bodies regardless of the posture of the spine. One of the topics crucial for better understanding of the stabilizing roles of spinal muscles but not investigated in the previous studies is the effect of strength of spinal muscles on the stability (or deflection) of the lumbar spine.

Computational analyses were conducted in this study using Wang's FE and optimization models in order to investigate the roles of spinal muscles in stabilizing the lumbar spine by creating the FCLs. For this purpose, the variation of spinal muscle strength was simulated by changing the values of MFCs of long muscles as well as of short intrinsic muscles (SIMs) as described in the Method section. Briefly, the strengthening (or weakening) of the muscle was modeled by increasing (or decreasing) the percentage ratio of MFC to average MFC of 45 N/cm^2 which is the median of possible MFCs (10 – 90 N/cm^2) [12, 24, 37, 41]. Five different MFC conditions (0%, 40%, 100%, 160%, and 200%) of both long muscles and SIMs in the spinal system in each postures were investigated in this study, and the trunk displacement (TD) and joint load including joint reaction forces (JRFs) and moments (JRMs) predicted from 25 cases of MFC variation were compared in order to investigate the effect of the strength of spinal muscles on the stabilization of the lumbar spine in a given posture. The results of TDs, total JRFs, and total JRMs predicted from 125 cases (5 postures x 125 MFC variations) showed significant effects of the spinal muscle strength on the stability and joint loads which vary in a posture-dependent manner as found in Tables 5-1 and 5-2.

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Small TDs were predicted from the spine in all postures when the strengths of both long muscles and SIMs were average or higher (MFC-L $\geq$ 100% and MFC-S $\geq$ 100%) as shown in Table 5-1. These results indicate that the lumbar spine can be stabilized by spinal MFs creating the FCLs along the path connecting the geometrical centers (GCs) of the adjacent vertebral bodies. Furthermore, minimal TD changes from the TDs of the average strength case (MFC-L = MFC-S = 100%) and total JRFs were found in all postures when the strength of both muscle groups increased (MFC-L and MFC-S >100%). These results indicate that the strengthening of spinal muscles above the average strength may result in no benefits in terms of stability improvement and/or less joint loads in the spine. Similar results were reported in previous studies of the relationship between the spinal stability and the forces in the long muscles [42, 43]. Stokes et al. found no significant changes in the spinal stability due to the weakening of long muscles up to 40% of maximum muscle contraction capability whereas no increase in the spinal stability due to the strengthening of long muscles was observed in another study. One of the possible physiological explanations for these results would be that spinal muscles are designed to have greater muscle capacity than the minimum requirement for maintaining the spine stably in a static condition. That is because, if the MFCs are equal to or less than its minimum requirement, there is no way to maintain the spinal stability when the MFCs are decreased by the muscle fatigue [44]. These results imply that it would be better to maintain the spinal muscle strength as high as possible in order to minimize the risk of back injury during a variety of activities by keeping the spine more stable.

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Table 5-1 Predicted TDs and the joint loads (JRFs and JRMs) in MFC variation of long muscles and SIMs. NS – Neutral Standing, FLX – Flexion, EXT – Extension, AR –Axial Rotation, LB – Lateral Bending

MFC-L	MFC-S	Trunk Displacements (mm)					JRF (N) / JRM (Nm)				
		NS	FLX	Ext	AR	LB	NS	FLX	Ext	AR	LB
0%	≥ 0%	NF (No FE predictions available)					NOS (No Optimum Solution)				
40%	0%	74.0	11.3	NF	118.4	NF	6092 / 16.2	9478 / 0	NOS	6036 / 59.1	NOS
	40%	1.0	8.0	104.0	120.0	NF	5245 / 0	8477 / 0	6313 / 61.7	6203 / 15.7	NOS
	100%	0.6	5.8	2.3	0.94	27.0	3971 / 0	7783 / 0	5855 / 0	4653 / 0	5104 / 57.1
	>100%	0.6	4.4	0.9	0.9	138.0	3861 / 0	7140 / 0	4497 / 0	4462 / 0	5402 / 40.1
100%	0%	1.3	2.2	18.7	7.4	2.1	6449 / 0	6001 / 0	10978 / 1.1	9328 / 4.9	5138 / 22.7
	40%	0.87	1.5	4.8	1.3	1.2	4766 / 0	5367 / 0	7138 / 0	5975 / 0	4141 / 17.2
160%	0%	1.3	0.9	22.0	3.2	0.6	6385 / 0	4564 / 0	8634 / 0	8905 / 0	3824 / 12.6
	40%	0.8	0.7	2.5	1.3	0.3	4748 / 0	4255 / 0	6570 / 0	5585 / 0	3480 / 5.4
200%	0%	1.3	0.9	20.0	2.9	0.5	6346 / 0	4564 / 0	8634 / 0	8546 / 0	3883 / 7.9
	40%	0.8	0.7	2.5	1.2	0.3	4748 / 0	4255 / 0	6560 / 0	5555 / 0	3970 / 1
≥ 100%	≥ 100%	0.7	0.9	1.6	1.8	1.2	3951 / 0	4880 / 0	4930 / 0	4247 / 0	5482 / 7.3

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In contrast, significant deficiency in the strength of both muscle groups (MFC-L $\leq$ 40% and MFC-S $\leq$ 40%) was found to result in either no-feasibility of optimum solutions or instability in a posture-dependent manner (Table 5-2). No optimum solutions of spinal MFs creating the FCLs in the spinal column were feasible with no contribution of long muscles (MFC-L = 0%) regardless of the strength of SIMs (MFC-S $\geq$ 0%) in all postures. These results indicate that the contribution of only SIMs is not sufficient to create FCLs in the spine because the bending moment produced by SIMs may not be large enough to compensate the large bending moment produced by the trunk weight. In fact, the bending moment produced by SIMs on the spine should be much smaller than that produced by the long muscles because of anatomical characteristics of SIMs, such as smaller PCSA (i.e., smaller maximum contraction force) and their location right next to the spinal column (shorter moment arms). Optimum solutions were also not feasible in the cases of extension and lateral bending postures with marginal contribution of long muscles (MFC-L=40%) but no SIMs (MFC-S=0%) as well as in the case of lateral bending posture with marginal contributions of both muscle groups (MFC-L = 40% and MFC-S = 40%). It was impossible to conduct the FE analyses for these cases of no feasible optimum solutions, and no model prediction data could be presented in Table 5.1.

The spinal instability (TDs > 5 mm) was also predicted in multiple cases with weak contribution of at least one muscle group (Table 5.2). In these cases, optimum solutions of MFs were feasible but the FE analyses with such MFs predicted large TDs and JRFs. It is interesting that the spine with no contribution of SIMs (MFC-S = 0%) was found unstable in extension posture regardless of the strength of long muscles (MFC-L > 40%) whereas the spine with weak contribution of long muscles (MFC-L = 40%)

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unstable in flexion and lateral bending postures regardless of the strength of SIMs (MFC-S $\geq$ 0%). These results indicate that the contribution of long muscles are more critical than that of SIMs in stabilizing the spine in flexion or lateral bending posture in which the spine is subjected to high bending moments whereas more contribution of SIMs are needed to stabilize the spine in extension posture. It was also possible to infer from Table 5.2 that it would be challenging to stabilize the spine in axial rotation posture with weak or no contribution of SIMs (MFC-S $\leq$ 40%).

The roles of SIMs in stabilizing the spine by MFs creating the FCLs could be found from the results of this study. The contribution of SIMs may not be essential for creating the FCLs in the spinal column when working with strong long muscles (MFC-L $\geq$ 0%). It would be necessary, however, not only to reduce the joint loads (JRFs and JRMs) significantly while keeping the stability but also to maintain the spine in more 3-D postures in a stable manner. The reduction of joint loads seemed to be achieved by allowing the use of smaller number of long muscles in creating the FCLs. SIMs are likely to provide more forces of small magnitudes in the directions which cannot be created by long muscles but are needed to stabilize the spine with smaller JRFs and JRMs. As such, the simultaneous use of SIMs and long muscles seems to be necessary to stabilize the spine in any physiological posture while minimizing the joint loads for maximum safety.

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Table 5-2 The cases of no optimum solution or unstable FE model in MFC variation of long muscles and SIMs. NS – Neutral Standing, FLX – Flexion, EXT – Extension, AR – Axial Rotation, LB – Lateral Bending

	MFC-L	MFC-S	Postures
No Optimum Solution	**0%**	**≥ 0%**	**All postures**
	40%	**0%**	**EXT, LB**
	40%	**40%**	**LB**
Unstable Cases	**40%**	**0%**	**NS, FLX, AR**
	40%	**40%**	**FLX, EXT, AR**
	40%	**100%**	**FLX, LB**
	40%	**>100%**	**LB**
	100%	**0%**	**EXT, AR**
	160%	**0%**	**EXT**
	200%	**0%**	**EXT**

Although the results of this study look reasonable and help us improving our understanding of the mechanics of the spine stabilization in-vivo using spinal muscles as previously discussed, but it is important to know the limitations of this study for correct understanding of the results. FE and optimization analyses were performed in this study using the numerical models of the spinal system that were validated with the experimental findings in the literature and used in a series of previous studies of the stabilization of the spine by spinal MFs creating FCLs [12, 24, 37]. While the details of model validations are well described in those previous studies, experimental investigations are required to find the agreement of the numerical predictions with in-vivo behaviors either directly or indirectly. Another limitation associated with the computational analyses of this study is the number of cases analyzed in this study. Continuous variation of independent variables, such as muscle strength and the posture of

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the lumbar spine, was not possible, and the computational analyses were performed on only 125 cases (5 long muscle group strengths $\times$ 5 SIM group strengths $\times$ 5 postures of neutral standing, flexion, extension, axial rotation, and lateral bending). Therefore, an example of distinct limitation would be that the least strength of long muscles for creating FCLs (MFC-L =40% equivalent to MFC-L =18 N/cm^2 may not be an exact threshold strength. A threshold strength smaller than 18 N/cm^2 could be available from the analyses of additional cases. However, it is not reasonable from a mechanical point of view to find a threshold strength value greater than 18 N/cm^2. Too many possible combinations for strength variations of 244 muscles made it impossible to study the effect of individual muscle strength variation. However, the results of this study may be good enough to investigate the stabilizing roles of long muscles vs. SIMs. The 5 postures considered in this study is also only a fraction of infinitely many postures made by the lumbar spine in-vivo. The postures investigated in this study represent the neutral standing as a reference posture and the extreme postures with physiologically possible maximum deflection and/or torsion. It is reasonable to expect the stabilization of the spine in posture within these maximum deformations of the lumbar spine using the MFs creating the FCLs and similar effects of the muscle strength variation as those found in this study. Another limitation is about the muscle forces. It is well known that a muscle force consists of two components, i.e., contraction force (active) and elastic force produced during the muscle elongation (passive). However, a muscle force predicted in this study represented the force along the muscle fascicle direction, which required to satisfying the given constraints with no distinction of active vs. passive portions. Such distinction, however, would not be necessary for the static analyses conducted in this

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study. As such, despite of these limitations, the results of this study are valid enough to provide meaningful guidelines for better understanding of a physiologically possible mechanism for the spinal stability provided by spinal MFs creating the FCLs. The results of this study should be helpful for finding physiologically normal biomechanics of the lumbar spine in-vivo.

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CHAPTER 6

CONCLUSION

The 3-D FE and optimization models of the lumbar spinal system developed in a previous study were used in this study with minor modification in order to investigate the roles of spinal muscles in stabilizing the lumbar spine by creating the FCLs as described in previous chapters [12, 24, 37]. The hypothesis tested in this study was that SIMs might play the major roles in stabilizing the spinal system while creating FCLs in the vicinity of the GCs of the vertebral bodies. For this purpose, the various conditions (0 – 90 N/cm^2) of MFCs of both long muscles and SIMs were simulated in five different postures, which are neutral standing, flexion 40°, extension 5°, left axial rotation 10°, and right lateral bending 30°. Throughout this study, it was found that small trunk sways (< 2mm) were predicted when MFCs of both long muscles and SIMs were average or higher regardless of the spinal postures. In contrast, no optimum solution or unstable conditions were predicted in many cases of the weakening of the long muscles, especially in flexion and lateral bending postures. Although the FCLs were created in most of the cases regardless of MFC-S when working with strong long muscles, higher joint loads were predicted as a result of weakening of SIMs. In addition, even if the long muscles were strong, absence of SIMs induced spine buckling in some cases of extension and axial rotation postures. The implication of these results can be summarized as follows:

- Human lumbar spine can be stabilized by spinal MFs creating the FCLs along the path connecting the geometrical centers (GCs) of the adjacent vertebral bodies.

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- The strengthening of spinal muscles above the average strength (45 N/cm^2) may result in no benefits in terms of stability improvement and/or less joint loads in the spine.
- The contribution of only SIMs is not sufficient to create FCLs in the spine regardless of the spinal posture.
- The contribution of long muscles are more critical than that of SIMs in stabilizing the spine in flexion or lateral bending posture in which the spine is subjected to high bending moments.
- The contribution of SIMs is more important to stabilize the spine in extension and axial rotation postures.
- Although the contribution of SIMs may not be essential for creating the FCLs in the spinal column when working with strong long muscles, it would be necessary not only to reduce the joint loads (JRFs and JRMs) significantly while keeping the stability but also to maintain the spine in more physiological 3-D postures in a stable manner.

These results partially support the hypothesis of this study for the reason that the stabilizing role of SIMs can be varied depending on the spinal postures, whereas contribution of the long muscles was required for FCLs creation regardless of the spinal postures. However, the concurrent use of both SIMs and long muscles is necessary for spine stabilization in any physiological posture with minimum joint loads for maximum safety.

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