BIOMECHANICAL AND PHYSICAL CHARACTERISTICS OF TRUNK AND HIP IN GOLFERS WITH AND WITHOUT LOW BACK PAIN

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Modified swing patterns and general exercises have been suggested for golfers with back problems. However, it is difficult to design a back-specific swing or exercise program for low back injury prevention and rehabilitation without knowing the differences in the kinematics and spinal loads of the golf swing and the physical characteristics associated with golfers with low back pain (LBP). The purpose of this study was to examine the kinematics of the trunk and spinal loads in golfers with and without LBP and their trunk and hip physical characteristics. Sixteen male golfers with a history of LBP were matched by age and handicap to 16 male golfers with no history of LBP. All golfers underwent a biomechanical swing analysis and physical characteristics assessment. Kinematics and spinal loads of the trunk were assessed using a 3D motion analysis system and two force plates. A bottom-up inverse dynamics procedure was used to calculate the spinal loads at L5/S1. In addition, trunk and hip strength and flexibility, back proprioception, and postural stability were measured. The LBP golfers demonstrated less trunk and hip strength. The LBP group also had less hamstring and right torso rotation flexibility. In addition, the LBP group demonstrated back proprioception deficits significantly in trunk flexion. No significant differences were found for postural stability. The LBP group demonstrated less maximum angular displacement between shoulders and hips during the backswing. No significant differences were found in other trunk kinematics and spinal loads during the golf swing. Deficits in physical characteristics have been found in golfers with a history of LBP. These differences may hinder dissipation of the tremendous spinal forces and moments generated by the golf swing over time and also limit trunk rotation during the backswing. These conditions may lead to lower back muscle strain, ligament sprain, or disc degeneration. Although differences found in this study can not be determined as causes or results of low back injuries in golfers, clinicians may be able to design an appropriate back-specific exercise program for golfers to prevent or rehabilitate low back injury based on these findings.

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PREFACE

Golfers are progressive athletes who are always seeking ways to improve their game. Golf is fun and enjoyable to study. However, without the support, encouragement, and assistance from the people around me, completion of this project would have been difficult.

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I. INTRODUCTION

Golf is both a sport and a recreational activity that can be enjoyed by people of all ages and skill levels. There are more than 55 million golfers around the world, according to an estimate in the early 2000s (Farrally, Cochran, Crews et al., 2003). With the increasing number of participants in golf annually, it is becoming popular worldwide (Batt, 1992; Gosheger, Liem, Ludwig et al., 2003; Sherman & Finch, 2000). Golf injuries and the athletic demands of the game are, therefore, more and more recognized because of the biomechanically complex golf swing. An inappropriate combination of muscle strength, flexibility, coordination, and balance (sub-optimal physical fitness) and an improper swing mechanics can cause abnormal stresses on the body that result in injury (Farfan, 1996; Hosea & Gatt, 1996; Watkins & Dillin, 1990).

A. Golf Injury to the Back

Epidemiologic studies have been conducted to identify injury risks patterns associated with golf. Among all golf-related injuries, low back pain (LBP) is the most common complaint (Lindsay, Horton, & Vandervoort, 2000; McCarroll, 1996; Sugaya, Tsuchiya, Moriya et al., 1999). Ninety percent of all Professional Golf Association (PGA) Tour injuries have been to the cervical and lumbar spine regions (Duda, 1987). During the 1990 competitive season, 59 % of the injuries reported on the PGA Tour involved the low back (Pink, Perry, & Jobe, 1993). Among amateur golfers, McCarroll et al. reported that 244 of 708 surveyed players suffered low back injuries (McCarroll, Rettig, &

Shelbourne, 1990). Batt, in a survey of 461 amateur golfers, found that 52% of men and 29% of women experienced back injuries (Batt, 1992). On average, low back injuries result in disability inhibiting golf participation for ten weeks (Gosheger et al., 2003). As such, it is essential to understand why low back injuries occur frequently among golfers.

In golf, the spine plays an important role in accomplishing the golf swing. It is involved with the transmission of forces and the coordination of activities between the upper and lower extremities. In sports requiring exertion of the upper extremities and trunk, such as golf, low back injuries are common (Farfan, 1996). However, in spite of the high incidence of low back injuries (Watkins & Dillin, 1990), scientific studies investigating the factors associated with low back injuries in golfers are limited.

B. Definition of Golf Related Low Back Pain

The golf swing produces considerable mechanical forces, including compressive force, shear force, and rotational force, to the lumbar spine due to rapid trunk bending and rotation (Hosea, Gatt, & Gertner, 1994). Improper swing mechanics and sub-optimal physical fitness can produce even larger and abnormal forces to the lumbar spine (Geisler, 2001; Hosea & Gatt, 1996; McCarroll & Gioe, 1982; Stover, Wiren, & Topaz, 1976). These forces often lead to the development of mechanical LBP which is generally localized to the lumbar area and associated with significant muscle spasms due to back muscle strain or spinal ligament sprain (Hosea & Gatt, 1996). It may begin gradually with episodes of exacerbation and result in permanent disability (Hosea & Gatt, 1996). Thus, participation in golf is increasingly being considered a mechanism for low back injuries. However, few studies have comprehensively investigated the differences in the lumbar mechanics and spinal loads during the golf swing and the differences in physical characteristics between golfers with and without mechanical LBP.

C. Mechanisms of Low Back Pain in Golfers

1. Kinematics and Kinetics of Golf Swing

X Factor and Spinal Loads

The golf swing consists of trunk rotations about three anatomical axes. During the backswing, the upper torso rotates against restricted pelvic rotation to produce maximum angular displacement between the shoulders and hips (the "X factor") (McLean, 1997). The maximum X factor usually happens at or shortly after the top of backswing when the pelvis starts to rotate forward leading into the downswing (unpublished data from the Neuromuscular Research Laboratory). This movement creates a tightly coiled body to store energy for maximum clubhead speed at impact. It also results in a large trunk rotation moment in the lumbar spine at the top of swing (Hosea & Gatt, 1996). If a golfer tries to generate a maximum X factor beyond physical limits of trunk rotation, an excessive rotational moment to the spine occurs that may not be well absorbed by the lumbar spine and may stress the soft tissue of the lumbar region over time. Lindsay and Horton observed this phenomenon (Lindsay & Horton, 2002). They found that the maximum X factors in the golfers with LBP and the golfers without LBP were similar. However, golfers with LBP tend to have less trunk rotational flexibility and as such the trunk rotation in their swings exceeds their physical limits.

Crunch Factor and Spinal Loads

In order to quantify lumbar spine mechanics that may result in low back injury, Morgan et al. proposed a parameter, the “crunch factor”, to describe the asymmetric movement during the golf swing (Morgan, Sugaya, Banks et al., 1997). The crunch factor is an instantaneous product of lumbar lateral bending angle and spinal rotation velocity. Morgan et al. hypothesized that the combination of lumbar lateral bending and high spinal rotation velocities during the downswing contribute to lumbar degeneration and injury as these fast movements cause a large lateral bending moment and a large rotation moment in the lumbar spine. Morgan et al. found that the maximum crunch factor occurs shortly after ball impact that was in agreement with many golfers’ subjective reports of pain just after impact (Morgan et al., 1997). In addition, Sugaya et al. observed that right-handed golfers with LBP showed greater right side degeneration at the facet joints and vertebrates compared to their left side (Sugaya, Moriya, Takahashi et al., 1997). Thus, Morgan et al. hypothesized that the crunch factor might be a useful measurement for assessing injury risk during the golf swing and can serve as the basis of comparison between ”healthy” and “pathological” golf swings.

In an attempt to test Morgan’s hypothesis, Lindsay and Horton calculated crunch factor by using the product of trunk lateral bending angle and spinal rotation velocity to compare the difference between golfers with and without LBP (Lindsay & Horton, 2002). Maximum trunk crunch factors of golfers with LBP were found to be smaller than that of golfers without LBP. Because of this reason, Lindsay and Horton indicated that if the whole trunk motion provides a representation of the lumbar crunch factor, then it would appear that factors other than the crunch factor must be responsible for the injury risks in golfers. This study tried to verify this controversy of the crunch factor by comparing the differences between golfers with and without LBP using exactly the same way described by Morgan et al.

Spinal rotation velocity and spinal loads

Spinal rotation velocity alone has been found to be significantly increased with low back injury risk (Marras, Lavender, Leurgans et al., 1995; Marras, Lavender, Leurgans et al., 1993). In golf, Hosea et al. have found that rapid spinal rotation velocity can produce considerable amount of spinal forces (Hosea, Gatt, Galli et al., 1990). Since tremendous spinal loads could result in low back injuries (Adams & Hutton, 1981; Farfan, Cossette, Robertson et al., 1970), the need to investigate the impact of spinal rotation velocity on the cause of LBP among golfers was warranted. Therefore, this investigation was to determine if spinal rotation velocity alone is a risk factor for back injuries during golf independent of trunk lateral bending.

Incline Factor and Spinal Loads

In addition to the traditional factors found in the literature related to LBP, a third factor, the "incline factor", was evaluated in this study. The incline factor is an instantaneous product of trunk flexion angle and lateral bending angle. In biomechanical analysis, the lever arm of the upper torso's center of mass increases due to trunk flexion, extension, and lateral bending that can increase the loads on the lumbar spine. In golf, trunk flexion and lateral bending angles are often emphasized from ball address to impact. Improper combination of these two angles during the golf swing can affect the shifting of the body center of mass within the base of support. This situation may limit the amount of trunk rotation available during the golf swing, cause more shear force in the spine, and increase the injury rate (Geisler, 2001). Thus, an inappropriate combination of trunk flexion and lateral bending was hypothesized to have a significant impact on the low back injuries by increasing spinal loads during the golf swing.

Reverse C Position and Spinal Loads

It is thought by many teaching and touring professionals that finishing the golf swing in trunk hyperextension ("reverse C" position) allows the golfer to well absorb the power released during the downswing, thereby increasing driving distance (Fischer & Watkins, 1996; Geisler, 2001). However, excessive extension of the spine can result in excessive anteriorly directed shear force on the lumbar spine (Geisler, 2001). Thus, this position is considered as a contributor to spinal injuries suffered by golfers. To finish a golf swing with the trunk in a more rotated and upright position is therefore suggested for golfers to prevent low back injuries (Fischer & Watkins, 1996). Whether golfers with LBP present greater trunk extension at the end of the golf swing than the golfers without LBP was examined in this study.

2. Physical Characteristics

Trunk and Hip Strength

Low back pain may be associated with the strength imbalance in trunk muscles that are responsible for the mechanical stability of the spine during activities (Andersson, Sward, & Thorstensson, 1988; Davis & Marras, 2000; J. H. Lee, Hoshino, Nakamura et al., 1999; McNeill, Warwick, Andersson et al., 1980). Trunk stability is usually created by the trunk flexors, extensors, and rotators that are responsible for the control of spinal orientation and load transfer across the spine (Bergmark, 1989). Weakness of any trunk muscle group would result in the inability of antagonist muscle group to generate enough force to counteract the moment produced by the agonist muscle, resulting in an instability of the spine (Davis & Marras, 2000; Panjabi, 2003). Research has attempted to identify the appropriate muscle strength ratios for trunk extension/flexion and rotation. There is a great variation in the assessment of trunk muscle strength due to the variety of testing methods. The most common cited strength ratio of extensor to flexor for healthy individuals is 1.3 : 1, indicating that the trunk extensors are 30% stronger than the trunk flexors (Beimborn & Morrissey, 1988). Ratios from 0.79 to 1.23 for extension/flexion have also been reported in the LBP patient population (Beimborn & Morrissey, 1988). Since the articulations of the back allow movements in multiple planes, it is important to look beyond just extension/flexion ratios and to identify the ideal ratio for trunk rotation. The generally accepted ratio for trunk rotation is 1:1 in healthy non-athletes (Beimborn & Morrissey, 1988). In healthy golfers, we have recently revealed this ratio of trunk rotation to be 1:1. However, it was not known if the golfers with LBP have different trunk strength ratios in extension/flexion and rotation compared to their healthy counterparts. This study investigated these differences to provide a basis for strengthening trunk muscles for injury prevention or rehabilitation.

In addition to trunk strength contributing to spinal stability, the association between the hip muscles and LBP is also important, as the hip muscles play a significant role in transferring forces from the lower extremities to the spine during ambulatory and sports activities (Lyons, Perry, Gronley et al., 1983; Nadler, Malanga, DePrince et al., 2000). Specifically, the hip abductor and adductor muscles are responsible for pelvic lateral stability (D. Lee, 1999). Side to side strength imbalance of the hip abductor and adductor muscles may cause pelvic obliquity and non-functional lumbar lateral bending (Kendall, McCreary, & Provance, 1993). Lumbar rotation coupled with non-functional lateral bending may overstress the soft tissue and facet joints around the lumbar spine, particularly during the golf swing (Davis & Marras, 2000; Hosea & Gatt, 1996). The gluteus maximus is a major pelvic stabilizer during trunk rotation (Lyons et al., 1983). Nadler et al. found a significant asymmetry in the strength of bilateral hip extensors in female athletes with reported LBP (Nadler, Malanga, Bartoli et al., 2002; Nadler et al., 2000; Nadler, Malanga, Feinberg et al., 2001). Furthermore, the iliopsoas muscles attach to the lumbar vertebrae. Although their major function is to control hip flexion movements, strength imbalance of bilateral iliopsoas muscles could potentially affect the bending of the lumbar spine during trunk and pelvic rotation resulting in LBP.

Under normal conditions, the trunk and hip muscles provide movement and stability to the trunk. Well-developed strength of the trunk and hip muscles is important for the athletes because most sports need relatively large movements of the trunk. However, unlike non-athletes whose back pain may result from weakness of the antagonist muscles, back pain in athletes may be caused by the increases of strength in the agonist muscles (Andersson et al., 1988). An accurate and objective assessment of the muscle strength based on a sport that an athlete plays is essential in injury prevention and rehabilitation. Little information about these issues exists for the golfers.

Trunk and Hip Flexibility

Trunk and hip flexibility deficits have been found to be associated with LBP in golfers (Vad, Bhat, Basrai et al., 2004). These deficits include decreased lumbar extension, decreased lead hip internal rotation, and increased FABERE's distance of the lead hip. FABERE's distance is the distance measured from the knee to the horizontal, with the subject in the supine position and the hip flexed, abducted, and externally rotated. A plausible reason for these deficits is that the lead hip acts as the primary pivot point while experiencing a significant amount of force that may lead to capsular contractures and subsequent rotation deficits over time. When the lead hip rotation is decreased, hip forces may be then transmitted to the lumbar spine and contribute to LBP. The limitation of lumbar extension may be due to a protective mechanism to decrease spinal loads and prevent further exacerbation of the LBP (Vad et al., 2004).

In the study of Vad et al., only trunk flexion, trunk extension, hip rotation, and FABERE's distance were measured. However, flexibility of trunk rotation (Mellin, 1987), hip extensor (Trainor & Trainor, 2004), hip flexor (Ashmen, Swanik, & Lephart, 1996; Mellin, 1988), hip abductor/adductor (Kendall et al., 1993), and hamstrings (Mellin, 1988) have also been considered to be associated with LBP in ordinary individuals. Tightness of these muscles may increase the stress placed on the spine by changing pelvic tilt and lumbar curve, resulting in low back injuries (Trainor & Trainor, 2004). Assessment of all trunk and hip flexibility mentioned above may be beneficial in evaluating the risk factors associated with golfer's back.

Back Proprioception

Proprioception is related to the sense of body position and movement. It helps the body to maintain stability and orientation during activities. Afferent input indirectly produces and modulates the efferent response that allows the neuromuscular system to maintain a balance of stability and mobility about joints (Laskowski, Newcomer-Aney, & Smith, 2000; Lephart & Fu, 2002). Poor proprioception can affect normal neuromuscular control, resulting in diminished joint stabilization, altered movement pattern, and a progressive decline of the joint (Lephart & Henry, 1992).

Deficient proprioception in back can be caused by back injuries, just as it is in peripheral joint injuries that the dysfunction of mechanoreceptors causes partial deafferentation of the joint (Laskowski et al., 2000; Yamashita, Cavanaugh, el-Bohy et al., 1990). Poor back proprioception may also contribute to the development of LBP. Dysfunction in the neural control system of the lumbar spine may place other spinal structures at risk for injury and also alter the spinal stabilizing system (Panjabi, 1992). Therefore, proprioception is considered a crucial element in the prevention and rehabilitation of recurrent spinal injuries, especially in sports. Few studies have reported that individuals with LBP present proprioception deficits in trunk flexion (Brumagne, Cordo, Lysens et al., 2000; Gill & Callaghan, 1998; Newcomer, Laskowski, Yu, Johnson et al., 2000). However, right-handed golfers with LBP symptoms exhibited a higher rate of right side vertebral and facet joint degeneration than non-golfing controls (Sugaya et al., 1999). It was not certain whether the asymmetrical lumbar spine injuries affect their proprioception in multiple trunk movement planes. Although many back rehabilitation programs have been designed to improve back proprioception under the assumption that back proprioception is lost in patients with back pain, the differences between golfers with and without LBP was not known and needed further examination.

Postural Stability

In addition to specific trunk measures, performance of the entire body should be considered to guide treatment and measure progress during recovery of LBP because trunk is the core of the kinetic chain of the human movement (Mientjes & Frank, 1999). Postural stability is easily disturbed in the presence of impairment in strength, coordination, and/or effective coupling of muscles in the lumbar and pelvic area (Luoto, Aalto, Taimela et al., 1998). Evaluation of the influence of LBP on balance control can be a measure of whole body performance and a suitable outcome measure of LBP (Mientjes & Frank, 1999). Mientjes & Frank found a significant increase in the root mean square (RMS) of the medial-lateral sway of the center of pressure (COP) in the patients with LBP while standing on both legs with both eyes closed (Mientjes & Frank, 1999). In a study of Luoto et al., sway velocity of the COP was also increased in the LBP group during one leg standing test with eyes open when compared to the healthy controls (Luoto et al., 1998). Based on these findings, similar patterns of poor postural control may exist in golfers with LBP that should be considered when back rehabilitation and injury prevention programs are designed.

3. Summary

Improper swing mechanics and sub-optimal physical fitness have been considered to be associated with the low back injuries in golfers. These factors may affect a golfer's back by increasing spinal loads during the golf swing. Swinging a club under sub-optimal physical fitness produces excessive forces to the lumbar spine, especially if combined with inappropriate swing mechanics. Maximum spinal loads during the golf swing are considered to be risk factors for causing low back injuries (Hosea & Gatt, 1996). However, the differences in the maximum spinal loads between golfers with and without LBP had not been examined. Thus, investigating swing mechanics and physical characteristics simultaneously with the maximum loads generated by their interactions in golfers with and without LBP may increase our knowledge regarding the mechanisms of low back injuries in golfers.

D. Statement of the Purpose

Modified swing patterns and exercise for golf have been suggested to reduce forces that create low back injury. However, without knowing the differences in the kinematics and kinetics of the golf swing and the physical characteristics between golfers with and without LBP, it is difficult to design an appropriate back-specific swing or exercise for low back injury prevention and rehabilitation. A comprehensive study of the factors associated with low back injury in golf was therefore necessary. The purpose of this study was to examine the kinematics and kinetics of the trunk in golfers with and without LBP and their physical characteristics, including trunk and hip strength, trunk and hip flexibility, back proprioception, and postural stability.

E. Specific Aims and Hypotheses

Specific Aim 1: To compare trunk kinematics between golfers with and without LBP during the golf swing. The comparison of the trunk kinematics included the maximum X factor normalized by the maximum trunk rotation angle toward the non-lead side in the neutral position, maximum crunch factor, maximum spinal rotation velocity, maximum incline factor, and the angle of trunk extension at the end of swing.

Hypothesis 1.1: The maximum X factor normalized by the maximum trunk rotation angle toward the non-lead side in neutral position would be larger in golfers with LBP compared to the golfers without LBP.

Hypothesis 1.2: The maximum crunch factor, maximum incline factor, and the angle of trunk extension at the end of swing would be larger in golfers with LBP compared to the golfers without LBP.

Hypothesis 1.3: The maximum spinal rotation velocity during the golf swing would be faster in golfers with LBP compared to the golfers without LBP.

Specific Aim 2: To compare the lumbar spinal kinetics between golfers with and without LBP during the golf swing. The comparison of the lumbar spinal kinetics included the maximum forces in the three anatomical axes (compression force, anterior-posterior shear force, and lateral shear force) and the maximum moments about the three anatomical axes (flexion-extension moment, lateral bending moment, and vertical rotation moment) at the L5/S1 joint.

Hypothesis 2.1: The maximum spinal forces at L5/S1 joint during the golf swing, including compression force, anterior-posterior shear force, and lateral shear force, would be greater in golfers with LBP compared to the golfers without LBP.

Hypothesis 2.2: The maximum spinal moments about L5/S1 joint during the golf swing, including flexion-extension moment, lateral bending moment, and vertical rotational moment, would be greater in golfers with LBP compared to the golfers without LBP.

Specific Aim 3: To compare trunk strength ratios and side-to-side hip strength differences between golfers with and without LBP. Trunk strength assessment included trunk flexion, extension, and right and left rotation. Hip strength assessment included bilateral hip abduction, adduction, flexion, and extension. Trunk strength ratios included trunk extension/flexion and right rotation/left rotation. Side-to-side strength difference of each hip muscle group was calculated as: absolute value of [(right hip strength – left hip strength)/maximum strength of both sides] x 100.

Hypothesis 3.1: Strength ratios of trunk extension/flexion and right rotation/left rotation would be different between golfers with and without LBP.

Hypothesis 3.2: Side-to-side strength difference of each hip muscle group, including hip flexors, extensors, abductors, and adductors, would be greater in golfers with LBP compared to the golfers without LBP.

Specific Aim 4: To compare trunk and hip flexibility between golfers with and without LBP. The measurement of trunk flexibility included trunk extension, flexion, and rotation. The measurement of hip flexibility included bilateral hip flexion, extension, abduction, adduction, internal rotation, external rotation, FABERE's distance and hamstring flexibility.

Hypothesis 4.1: Trunk flexibility, including trunk extension, flexion, and rotation would be less in golfers with LBP compared to the golfers without LBP.

Hypothesis 4.2: Average of bilateral hip flexion, extension, and hamstring flexibility that affects the lumbar spinal curve in the sagittal plane would be less in golfers with LBP compared to the golfers without LBP.

Hypothesis 4.3: The flexibility of hip abduction, adduction, internal rotation, external rotation, and FABERE’s distance would be less on the lead side compared to the non-lead side within golfers with LBP. There would be no differences between legs in these comparisons within golfers without LBP.

Specific Aim 5: To compare back proprioception between golfers with and without LBP. Active spinal repositioning error was measured as a method of measuring back proprioception. Three-plane movement, including flexion/extension, bilateral side bending, and rotation were tested.

Hypothesis 5.1: Active spinal repositioning error in the sagittal plane, including flexion and extension would be larger in golfers with LBP compared to the golfers without LBP.

Hypothesis 5.2: Active spinal repositioning error in the frontal plane, including right and left side bending would be larger in golfers with LBP compared to the golfers without LBP.

Hypothesis 5.3: Active spinal repositioning error in the horizontal plane, including right and left rotation would be larger in golfers with LBP compared to the golfers without LBP.

Specific Aim 6: To compare the single-leg standing balance between golfers with and without LBP. The comparison included the sway velocity of the COP while standing on one leg with eyes open and eyes closed.

Hypothesis 6: The sway velocity of the COP while standing on one leg with eyes open and eyes closed would be faster in golfers with LBP compared to the golfers without LBP.

II. REVIEW OF LITERATURE

A. Biomechanical Analysis of Trunk Motion in Golfers

Trunk motion plays an important role in the development of low back injuries, particularly when motion occurs simultaneously in multiple planes (Davis & Marras, 2000). The golf swing involves multiplanar rapid movements of the trunk. Low back injury is reported to be the most common complaint of golfers. Therefore, it is necessary to understand the impact of trunk motion on golfers' backs in order to reduce the risk for low back injuries.

1. Kinematics and Kinetics of Trunk Motion During the Golf Swing

X factor and spinal loads

Lindsay and Horton compared trunk motion during the golf swing between six professional golfers with LBP and six professional golfers without LBP (Lindsay & Horton, 2002). Golfers with LBP had less trunk rotational flexibility in the neutral standing position. This resulted in spinal rotation beyond their physical limits during the backswing in order to achieve a position with more coil for downswing. The maximum trunk right rotation angles during the backswing were similar between the back pain group (34.8 ± 7.3°) and the pain free group (35.6 ± 4.2°). After these values were normalized by the maximum trunk right rotation angle in neutral posture, the percentages were 108.3 ± 20.0 % and 88.0 ± 24.9 % in the LBP and pain free group, respectively.

Although the difference of this comparison was not statistically significant due to the small sample size, Lindsay and Horton suggested that players with LBP should improve their trunk rotation flexibility to reduce the stress that may occur on the lumbar vertebrates during the backswing.

Crunch factor and spinal loads

Sugaya et al. conducted a radiographic study on right-handed amateur and professional golfers with LBP (Sugaya, Moriya, Takahashi et al., 1996). Golfers with LBP had greater right side degeneration at the lumbar facet joints and vertebrae in comparison to the age-matched non-golfers with LBP. Based on this evidence, Morgan et al. proposed the "crunch factor", the product of lumbar lateral bending angle and rotation velocity, to examine the kinematics of the lumbar spine during the golf swing in an attempt to compare "healthy" and "pathological" golf swings (Morgan et al., 1997). These authors tested ten healthy right-handed male collegiate golfers with the mean handicap of 3.2 ± 3.4. The results revealed that the peak value of the crunch factor occurred about 52 milliseconds after impact. Lumbar axial rotation velocity reached its greatest value around 25 milliseconds after impact. Morgan et al. (1997) indicated that these findings correlate well with the reports that greatest back pain occurred shortly after impact. Thus, they believe that the crunch factor might be a useful measurement for assessing injury risk during the golf swing. However, unlike maximum spinal forces that were observed prior to or at ball impact (Hosea et al., 1990), the maximum crunch factor happened after ball impact. The timing of the maximum crunch factor is not the same as that of the maximum spinal loads which are thought to be a cause of low back injury. If the crunch factor does not correlate well with the spinal loads, it may not be a stronger contributor to LBP.

Lindsay and Horton (2002) tried to determine if the crunch factor proposed by Morgan et al. (1997) is an important contributor to low back injury. Because the method that Lindsay and Horton used did not allow lumbar motion to be isolated from thoracic motions, they calculated overall "trunk crunch factor" by using the product of whole trunk lateral bending and axial rotation velocity. Lindsay and Horton then compared the maximum trunk crunch factors between the golfers with and without LBP. The maximum trunk crunch factor in the LBP group (82.4 ± 21.9 rad/s) was smaller than that of the group without LBP (87.7 ± 28.4 rad/s). This finding was against the hypothesis of Morgan et al. (1977). Lindsay and Horton then pointed out that factors other than the crunch factor may be able to better describe the differences in swing mechanics between golfers with and without LBP if the trunk crunch factor can be a representation of the lumbar crunch factor. Based on this argument, further verifying the hypothesis of the crunch factor by using the method of Morgan et al. and also looking for another potential indicator for the cause of LBP in golfers are both important and should be considered.

Spinal rotation velocity and spinal loads

Rapid spinal rotation velocity during the golf swing has been reported to be a risk factor for low back injuries (Hosea et al., 1990). Increased rotation velocities would result in increased lumbar spinal forces. Lindsay and Horton (2002) measured trunk rotational velocities between golfers with and without LBP. Trunk rotational velocities during the downswing in the back pain group (3.25 ± 0.58 rad/s) were not significantly faster than that of the pain free group (3.18 ± 1.62 rad/s). The majority of the spinal rotation occurred via the thoracic spine. Whether the rotational velocities of the lumbar spine are different between golfers with and without LBP was not known. Since the low back injuries in golfers are thought to be caused by the forces that are associated with the lumbar movements, it is necessary to investigate the lumbar spinal rotation velocities separately from the whole trunk rotation velocities.

Trunk tilt and spinal loads

In an effort to increase clubhead speed, many golfers improperly use an excessive lateral weight shift of the lower body at the top of the backswing, rather than the proper mechanics of rotating the pelvis (Geisler, 2001). When the lower body is shifted to the right, the spine bends laterally to the left in order to maintain balance. This compensatory movement will result in difficulty for golfers to retain the anterior tilt angle of the spine that they had at address position (Geisler, 2001). A golfer using this swing pattern will be forced into reverse trunk inclination in the early downswing period by aggressively sliding the hips back laterally toward the target in order to hit the ball. This chain reaction forces the spine to simultaneously bend laterally to the right in order to reestablish the original spine and torso inclination. When this violent loss and reestablishment of the spine angle occurs, the spine undergoes significant rotation and shear forces (Geisler, 2001). This potential injury mechanism may be able to explain why golfers who demonstrated greater left side bending on the backswing (6.7 ± 3.2° versus 0.5 ± 3.1°) had LBP resulting from the golf swing (Lindsay & Horton, 2002).

Finishing the golf swing in trunk hyperextension has been considered as a risk factor to the low back injuries caused by increased spinal forces. However, studies had not examined if golfers who experienced LBP hyperextend their trunk more than those golfers who never had back pain from golf. Lindsay and Horton (2002) demonstrated that the maximum trunk extension angles during the golf swing in golfers with LBP (-2.3 ± 8.5°) were less than that of the golfers without LBP (-10.2 ± 8.0°). This may be a protective mechanism adopted by injured golfers to prevent back pain since they were tested with existing pain. This study was verify if trunk hyperextension in the end of the golf swing contributes to the development of LBP.

2. Trunk Muscle Activities During the Golf Swing

Hosea et al. investigated trunk muscle activity together with the spinal loads during the golf swing on right-handed professional and amateur golfers (Hosea et al., 1990). Muscle activities measured in this study included bilateral erector spinae, rectus abdominis, and external oblique. Overall, the muscles on the left side of the trunk were responsible for the initial twisting of the trunk from address to the top of backswing. These muscles produce lumbar axial torque to the right during the backswing period. The muscles on the right side of the trunk, on the other hand, play a major role during the downswing and created axial torque to the left. During the downswing, maximal activities of all trunk muscles occurred (particularly the right side muscles) which corresponded to the maximum spinal loading of the anterior-posterior shear force, lateral shear force, and axial torque. Back muscles, however, contracted symmetrically during this period for stabilizing the spine and generated compression force.

Pink et al. analyzed the muscle firing patterns in the trunk during the golf swing

(Pink et al., 1993). Muscle activities of bilateral erector spinae and abdominal oblique were sampled at 2500Hz and recorded on 23 right-handed golfers with handicaps of five or below. A high-speed motion picture camera (400Hz) positioned in front of the subjects recorded their swing performance. Electronic marks were placed on the film and EMG data to allow for synchronization. The motion was divided into five segments: 1) take-away: from ball address to the end of the backswing, 2) forward swing: from the end of the backswing until horizontal club, 3) acceleration: from horizontal club to ball contact, 4) early follow through: from ball contact to horizontal club, 5) late follow through: from horizontal club to the end of motion. The results demonstrated relatively low activity in all muscles (below 30% of maximum voluntary contraction, MVC) during take-away. From forward swing to early follow through, relatively high and constant activity was observed in all muscles (above 30% of MVC). Right abdominal oblique and erector spinae muscles were particularly active during the trunk rotation in the forward swing and acceleration phases. Pink et al. indicated that back muscles may primarily contract for trunk stabilization and the abdominal muscles may contract for trunk flexion and rotation. These muscles functioned to initiate and control the trunk movements, transmit the power initiated from the hips, and decelerate the body after ball impact. Thus, Pink et al. believe that a strengthening program for trunk muscles is needed for golfers to prevent injuries from overuse and abnormal motions from the poor trunk mechanics.

Watkins et al. conducted a study similar to the study of Pink et al. (1993) on thirteen right-handed professional male golfers (Watkins, Uppal, Perry et al., 1996). Muscle activities were collected on bilateral abdominal obliques, upper rectus abdominis, lower rectus abdominis, erector spinae, and gluteus maximus during the golf swing. The patterns of bilateral trunk muscle activities found by Watkins et al. were the same as that observed by Pink et al. (1993). Additionally, Watkins et al. noticed that the gluteal muscles were very active during the forward swing and acceleration phase. This indicates the role of the hip stabilizers and the initiation of power to start the drive of the golf club into the acceleration phase (Watkins et al., 1996). The results of Watkins et al. study demonstrated that the strength of trunk and hip muscles is important and the coordination of these muscles is essential for the golfers. Watkins et al. also speculated that the patterns of trunk muscle activity may be different between golfers with and without LBP.

Horton et al. compared abdominal muscle activation patterns between elite golfers (professional or with handicap less than 5) with and without LBP (Horton, Lindsay, & Macintosh, 2001). Surface EMG data were sampled at 2400 Hz and collected bilaterally from the rectus abdominis, external oblique, and internal oblique muscles. Four high-speed video cameras (240 Hz) were synchronized with EMG data to determine the phases of golf swing. The phases of golf swing were defined similar to the study of Pink et al. (1993). The results demonstrated no differences between groups in the root mean square (RMS) of abdominal muscle activities during the golf swing. However, the onset times of the external oblique muscles on the lead side were significantly delayed (40ms) with respect to the start of the backswing in the back pain group. Abdominal muscle fatigue, as measured with median frequency and RMS, did not show significant difference between the two groups after 50-minute practice session. Significant differences in the onset times of external oblique (lead side) muscle activity between golfers with and without LBP may suggest inappropriate recruitment of these muscles in golfers with LBP during the golf swing (Horton et al., 2001). This difference may also relate to the deficits of the neuromuscular control. However, Horton et al. indicated that injured golfers were tested with existing chronic LBP. The deficits shown in this study could not be concluded as causes or results of the pain. Horton et al., therefore, suggested that future studies should use other ways to assess trunk muscle endurance or investigate other possible causes of LBP among golfers.

3. Biomechanical Models for the Spinal Load During the Golf Swing

Biomechanical models have been used to estimate spinal loads and identify high-risk movements and activities for several decades. This approach usually requires many simplifying assumptions for the properties and structures of the mechanical system to determine equations for a mathematic model. Depending on the need of the research purpose and the complexity of the motion, linked segment models (LSMs) using an inverse dynamics approach or combined with the EMG signals from the trunk muscles were adopted for appropriate estimation for the spinal loads in dynamic analysis. However, most of the models were used for lifting tasks to investigate low back injuries in industry. Only a few attempts were made to estimate the lumbar spinal load during sports activities. This may be due to the complex and fast body movements in sports resulting in the difficulty of obtaining or controlling necessary variables for the calculation.

Hosea and associates attempted to identify the lumbar spinal load at the L3-4 spinal motion segment during the golf swing (Hosea et al., 1990). They used a model modified from Cappozzo's technique (Cappozzo, 1983). Cappozzo used top-down inverse dynamics approach combined with estimated trunk muscle forces to develop a 3D dynamic biomechanical model for spinal load estimation at L3-4 level during level walking in a straight line. The model included four segments of the upper body, including the head, upper torso, and the two upper extremities. Hosea et al. modified the two upper extremities used in Cappozzo's model into four separate upper and lower arm segments, allowing them to take into account the movements of elbow joints during the golf swing for the estimation of spinal loads. Overall, this model summed the gravitational, inertial, and muscle forces to determine the total loads at the L3-4 motion segment. These spinal loads included anterior-posterior shear force, compression force, lateral bending force, and axial torque.

Lim and Chow estimated the loads acting on the L4-5 spinal motion segment during the golf swing using an EMG-assisted optimization model developed by Cholewicki and McGill in 1994 (Lim & Chow, 2000). This model incorporated the muscle recruitment patterns while satisfying the equations of moment equilibrium. In addition to the middle and lower trunk segments, the body segments included in Lim and Chow's study (2000) were bilateral thighs, lower legs, and feet. The spinal loads were calculated using bottom-up inverse dynamics approach with muscle forces estimated from twenty-two trunk muscles. Using this model, Lim and Chow estimated the spinal loads during the golf swing, including anterior-posterior shear force, lateral shear force, and compression force.

The models of Hosea et al. (1990) and Lim et al. (2000) provided valuable information regarding the spinal loads during the golf swing for identifying the potential risks for vertebral disc injury. Due to the difference in the various modeling techniques, the amplitudes and patterns of the spinal forces in each movement direction are not identical between the studies of Hosea et al. and Lim et al. Hosea et al. noted that the large compressive force (8 times of body weight) generated during the golf swing may fracture the intervertebral disc and the pars interarticularis. Lim and Chow, who found that the maximum compressive force was 7 times of body weight, concluded that the magnitude of the load may not be the primary factor for causing low back injury. The accumulated stress due to repeated golf swings could be a major reason for disc degeneration and LBP in golfers (Lim & Chow, 2000).

Although the amplitude of spinal loads has been found to be different between models used for the calculation, these loads may predispose golfers at any skill level to developing muscle strains, vertebral disc injury, spondylolythesis, and facet joint arthritis. In addition to improving swing mechanics, a comprehensive conditioning program should be emphasized for all golfers in order to reduce spinal load during the golf swing and protect them from back injuries.

B. Physical Characteristics and Low Back Pain

In addition to trunk dynamics, muscle strength and flexibility of trunk and hip, back proprioception, and postural stability are also essential factors contributing to the relationship between trunk motion and low back injuries. Studies have investigated the differences of these physical characteristics between individuals with and without LBP, although little of this information was collected specifically from golfers. An understanding of the role of trunk and hip strength, flexibility, back proprioception, and postural stability in LBP would aid in designing specific injury prevention or rehabilitation programs for golfers.

1. <u>Muscle Strength in Low Back Pain</u>

Trunk and hip muscles are considered core muscles because they work together to stabilize, move, and protect lumbo-pelvic-hip complex during functional activities (Clark, 2001). Lack of strength in these muscles will decrease the ability to produce efficient movements, which could lead to low back injury (Hodges & Richardson, 1996). Therefore, strengthening of trunk and hip muscles is usually one of the major exercise programs for the prevention or rehabilitation of LBP. Meanwhile, strength imbalances of these muscles has also been put forward as possible factors in the etiology of LBP (McNeill et al., 1980; Nadler et al., 2001). However, specific information in regard to the prevention of LBP in golf had not been established. An accurate assessment of the trunk and hip muscle strength is necessary for this population because it can help to indicate individual golfer's functional capacity and prevent injury.

Trunk Muscle Strength

Strengthening of the muscles which support the lumbar spine is commonly recommended for patients with LBP as a method of treatment, as well as for the healthy population as a possible preventive measure for LBP (Gundewall, Liljeqvist, & Hansson, 1993). However, studies have shown inconsistent results in the relationship between LBP and trunk muscle strength. Some investigators have reported that the maximum voluntary contraction strength in individuals with LBP is weaker than that of the healthy controls (Mayer, Smith, Keeley et al., 1985). Others have found no difference in trunk strength between groups (Newton, Thow, Somerville et al., 1993). Whether trunk muscle weakness contributes to the incidence of LBP is also controversial. Some studies have revealed a negative correlation between trunk muscle strength and the incidence of LBP (Biering-Sorensen, Thomsen, & Hilden, 1989; Chaffin, Herrin, & Keyserling, 1978). It has also been pointed out that weak trunk muscle strength is one of the strongest risk indicators for a first-time experience of LBP (Biering-Sorensen et al., 1989). However, other researchers disagreed with the above findings because none of their isokinetic measurements showed significant differences between individuals who developed LBP and those who did not develop LBP (Mostardi, Noe, Kovacik et al., 1992; Newton et al., 1993). These authors then concluded that trunk muscle weakness did not correlate with the incidence of LBP.

Despite these apparent contradictions, a study by Lee and coworkers provided further information on whether trunk weakness is a risk factor for LBP (J. H. Lee et al., 1999). In this 5-year prospective study, trunk muscle strength was measured isokinetically (60 degrees/sec) from individuals who neither reported nor had ever been treated for LBP. The peak torques of the trunk extension, flexion, right rotation, and left rotation were measured using a trunk extension/flexion unit and a torso rotation unit. The agonist/antagonist ratios were calculated as extension/flexion and left rotation/right rotation ratio. The subjects then were followed for 5 years to determine the incidence of LBP and were classified into a non-LBP group (subjects with no LBP during the 5-year follow-up period) and a LBP group (subjects who experienced LBP during this period). The results revealed no significant differences between groups regarding age, height, weight, the peak torque values, or the left rotation/right rotation ratio. However, the extension/flexion ratio of the LBP group (men, 0.96 ± 0.27; women, 0.77 ± 0.19) demonstrated significantly lower values than that of the non-LBP group (men, $1.23 \pm$

0.28; women, and 1.00 ± 0.16). The authors concluded weaker trunk extensors than flexors may be a risk factor for LBP. The results of this study also showed similar extension/flexion strength ratios to that measured directly from patients with LBP in previous studies. The range of extension/flexion strength ratios found in previous studies was 1-2:1 and 0.79-1.23:1 in the healthy and back pain groups, respectively (Beimborn & Morrissey, 1988).

Trunk rotation has been mentioned as a potential risk factor associated with LBP (Manning, Mitchell, & Blanchfield, 1984; Marras et al., 1993). An epidemiological study conducted by Manning et al. (1984) demonstrated that 11.4% of accidental back injuries and 49% of overuse back injuries involved rotation or twisting of the trunk. Trunk rotation was also found to be the second most hazardous motion for low back disorders in repetitive industrial lifting jobs (Marras et al., 1993). Having strong trunk rotation strength to overcome the passive resistance produced by the movements with trunk rotation may be able to protect the back from injury. Studies have found a strength ratio of 1:1 between left and right trunk rotation in healthy individuals (Kumar, Dufresne, & Van Schoor, 1995; J. H. Lee et al., 1999; Toren & Oberg, 1999). The authors of the current study have also revealed same strength ratio of trunk rotation in healthy golfers recently. Golfers need fast and powerful trunk rotation during the golf swing. It would be helpful to examine the ratio of trunk rotation strength between golfers with and without LBP and to identify if this is a risk factor for LBP.

Hip Muscle Strength

Hip muscles are responsible for pelvic stability and force transmission between the lower extremities and trunk (D. Lee, 1999; Lyons et al., 1983). Thus, it is suggested that these muscles need to be trained in addition to the muscles attached to the lumbar spine directly, in order to provide sufficient pelvic and trunk stability (Clark, 2001). Kankaanpaa et al. found that individuals with LBP demonstrate poor endurance of the gluteus maximus muscles (Kankaanpaa, Taimela, Laaksonen et al., 1998). This decreased endurance may be the result of disuse related to the low back injury, or it may be the cause of the injury. In the cohort study by Nadler et al., a significant difference in side-to-side symmetry of maximum hip extension strength was observed in female athletes who reported LBP during the previous year as compared to those who did not (Nadler et al., 2000). Female athletes with LBP had left hip extensors that were 15% stronger than their right extensors. This difference was only 5.3% (left side stronger) in the group without LBP. A subsequent prospective study by Nadler et al. that assessed whether athletes with strength imbalance of the hip muscles would be more likely to require treatment over the following year supported the results of the previous cohort study (Nadler et al., 2001). It was reported that the percentage of strength difference between the right and left extensors in female athletes was predictive of whether treatment for LBP was required over the ensuing year. Hip flexor strength deficits may also be related to LBP. Decreased hip flexor strength was found in patients with LBP (J. H. Lee, Ooi, & Nakamura, 1995). Weakness or imbalance of the iliopsoas, the major hip flexor, may affect lumbar lordosis and result in LBP (Kisner & Colby, 1990).

The hip abductors stabilize and prevent a downward inclination of the pelvis (Trendelenburg sign) during single leg stance. If hip abduction strength is weak, the requirements of the lateral trunk stabilizers, such as the quadratus lumborum, will be increased in order to better stabilize the pelvis (Nadler et al., 2002). A muscle imbalance may decrease normal lumbo-pelvic-hip stability and contribute to the low back injury. In addition, hip abductors and adductors help to maintain the sacroiliac joint stability, thus disturbance of their function may result in sacroiliac joint instability and lead to LBP (D. Lee, 1999). Weakness of hip abductors and adductors and asymmetrical hip abductor and adductor strength may be associated with LBP (Nadler et al., 2002; Nourbakhsh & Arab, 2002). Whether these conditions also exist in golfers with LBP was examined in this study.

Trunk and Hip Strength in Athletes

Imbalances in trunk and hip muscle strength may be a factor causing LBP, as mentioned above (McNeill et al., 1980; Nadler et al., 2001; Thorstensson & Arvidson, 1982). The abnormal ratios in non-athletes with LBP may be due to specific weakness in hip muscles or trunk extensors. However, the abnormal ratios in the athletes may be caused by specific increase in the strength of muscles acting in the opposite direction (Andersson et al., 1988). Selective increase of the strength of certain trunk and hip muscles in the athletes appears to be related to long-term systematic training of a specific sport (Andersson et al., 1988), resulting in agonist and antagonist strength ratios that are different from the ratios of healthy individuals who are not athletes.

The muscles of the trunk and hips play an important role in performing efficient movements and maintaining spinal stability. Strengthening of these muscles is necessary for controlling the body movements in many sports, especially for those sports requiring large and fast movements of the trunk. These great demands of the trunk will be exerted on the trunk and hip muscles since the trunk segment has a large mass (Andersson et al., 1988). To design an effective injury prevention or rehabilitation program for athletes, it is important to understand how a specific muscle group functions for a sport and also to pay attention to the muscle balance with its antagoist muscle group. Thus, the risk of injury can be diminished and the athletes with LBP can resume their normal functional activities soon and also prevent re-injury.

2. Flexibility in Low Back Pain

Appropriate flexibility of trunk and hip muscles is suggested to decrease risks of low back injuries in golfers (Lindsay & Horton, 2002; Vad et al., 2004). Appropriate flexibility can decrease the resistance in various tissues in the trunk and hip muscles and prevent the changes of normal lumbar curve caused by tight muscles. A golfer with appropriate flexibility is therefore less likely to incur injury by exceeding tissue extensibility during the golf swing. This may then decrease the forces applied to the lumbar spine.

Vad and colleagues assessed trunk and hip flexibility in professional golfers by measuring the range of motion (ROM) of lumbar flexion, lumbar extension, bilateral hip rotation and the distance from knee to the horizontal table while performing FABERE's test (hip flexed, abducted, and externally rotated) (Vad et al., 2004). Fourteen golfers with a history of LBP for more than 2 weeks within the year prior to their measurements were compared to 28 control subjects. The results revealed that golfers with previous LBP had less lumbar extension angles than the control subjects. The LBP group also had flexibility deficits in the hip internal rotation angle and FABERE's distance of the lead leg when compared to their non-lead legs. In the study of Lindsay and Horton, trunk rotation during the golf swing in golfers with LBP was observed to exceed their maximum voluntary trunk rotation in neutral posture (Lindsay & Horton, 2002). However, golfers without LBP rotated trunk within their maximum rotation range in neutral posture. Since there were no differences in the trunk rotation angle between golfers with and without LBP before normalizing to their maximum voluntary trunk rotation in neutral posture, the results of the Lindsay and Horton's study imply that golfers with LBP need to increase their trunk rotation flexibility.

In addition to the flexibility investigated by Vad et al. and Lindsay and Horton, tight hip extensors and hamstrings may decrease lumbar lordorsis and tight hip flexor may increase lumbar lordosis (Neumann, 2002a, 2002b). An abnormal lumbar posture will place extra stress on the spine (Trainor & Trainor, 2004). Studies have demonstrated a positive association between LBP and decreased flexibility of hip extensors, hip flexors and hamstrings (Ashmen et al., 1996; Mellin, 1988). Similarly, tightness of hip abductors and adductors can limit the motion of pelvis or even cause pelvic lateral tilt if both sides are not symmetrical (Kendall et al., 1993). This will then affect functional lumbar lateral bending during the activities, and result in increased spinal forces.

Flexibility is an important component of physical fitness. The requirements of flexibility are sports-specific and joint specific. Athletes need to have sufficient musculoskeletal flexibility to meet the demands of their sport. Otherwise, performance will be sub-optimal and the risk of injury will be increased (Gleim & McHugh, 1997). In order to improve golfers performance and prevent low back injuries, all potential flexibility deficits around the lower back in golfers with LBP need to be evaluated.

3. Proprioception in Low Back Pain

Studies have shown proprioception deficits in injured joints, such as the shoulder (Lephart, Myers, Bradley et al., 2002; Warner, Lephart, & Fu, 1996), knee (Borsa, Lephart, & Irrgang, 1998; Borsa, Lephart, Irrgang et al., 1997; Lephart, Giraldo, Borsa et al., 1996; Lephart, Pincivero, Giraldo et al., 1997; Safran, Allen, Lephart et al., 1999; Simmons, Lephart, Rubash, Borsa et al., 1996; Simmons, Lephart, Rubash, Pifer et al., 1996), and ankle (Rozzi, Lephart, Sterner et al., 1999). Similarly patients with LBP also demonstrated difficulty in adopting and returning to a neutral position of the lumbar spine or a position in the middle of the lumbar ROM (Gill & Callaghan, 1998; Newcomer, Laskowski, Yu, Johnson et al., 2000; Parkhurst & Burnett, 1994). These proprioception deficits may influence the motor programming for neuromuscular control and muscle reflexes that provide dynamic joint stability (Lephart et al., 1997). Meanwhile, ligamentous trauma may cause mechanical instability and proprioceptive deficits that will disturb functional stability and lead to further micro-trauma and re-injury (Lephart et al., 1997). It is important for an injured athlete to regain neuromuscular control before returning to competition. If proprioception training can be integrated early into a training program, athletes will be able to have significant improvement in functional and sport-specific activities soon after injury and rehabilitation (Lephart et al., 1997).

Mechanoreceptors that are responsible for proprioceptive input have been found in a number of spinal connective tissues, facets joints, and discs (McLain & Pickar, 1998; Roberts, Eisenstein, Menage et al., 1995). There are four types of mechanoreceptors: Ruffini endings, Pacinian corpuscles, Golgi tendon-like organs, and free nerve endings (Grigg, 1994; McLain & Pickar, 1998). Each of these responds to different stimuli and gives specific afferent information that modifies neuromuscular function for providing proprioceptive sense, modulating protective muscular reflexes, and signaling tissue damage while performing excessive movement (Grigg, 1994; McLain & Pickar, 1998). However, the number of receptive endings in the lumbar facet capsules was found to be small (McLain & Pickar, 1998). It was suggested that these receptors may have a relatively large receptive field. One or two nerve endings may be sufficient to monitor the area of each facet joint. Damage to even a small area may denervate the facet and have important implications for long-term spinal joint function (McLain & Pickar, 1998).

Static stability of the lumbar spine is maintained by the bony and ligamentous structures while dynamic stability is supported by trunk muscles, particularly the deep muscles that attach directly to the lumbar vertebrates (Cholewicki & VanVliet, 2002; Ebenbichler, Oddsson, Kollmitzer et al., 2001; Lam, Jull, & Treleaven, 1999; Panjabi, 1992). Patterns of trunk muscle activation are usually investigated with the help of electromyography. It was suggested that muscle recruitment is controlled by the central nervous system according to the task to be performed (Granata & Marras, 1995). Studies have found the changes of muscle contraction patterns in patients with LBP (Hodges, 2001; Hodges & Richardson, 1998, 1999). These changes reflect different motor control strategies between individuals with and without LBP. Appropriate timing and coordination of trunk muscle contractions are required for maintaining normal dynamic joint stability. However, inefficient muscular stabilization of the lumbar spine has been observed in patients with LBP (Hodges & Richardson, 1996, 1997). Proprioception modulates the afferent input and efferent response that allows the neuromuscular system to maintain a balance and mobility (Laskowski, Newcomer-Aney, & Smith, 1997). It is important to maintain a good proprioception for coordinating trunk muscles and controlling human movements.

Many studies have assessed back proprioception in individuals with and without LBP. Several of these studies have found a relationship between back proprioception and low back injury, although others were not successful due to the difficulty of the measurement technique. Based on the results of these studies, methods that can improve the measurement of back proprioception will be discussed in the section of methodological consideration in this chapter. Even though the measurement of back proprioception is not easy, Parkhurst and Burnett were able to demonstrate a relationship between low back injury and altered proprioception of the low back in a group of 88 firefighters (Parkhurst & Burnett, 1994). These authors examined three types of low back proprioception: passive motion threshold, directional motion perception, and repositioning accuracy. Three planes of spinal motion were measured for each type of proprioception using a custom designed spinal motion apparatus composed of several tables and seats that allowed upper body to be fixed on a support. A continuous passive motion machine was connected to a moveable table or chair for moving the lower body. Low back injuries were correlated with proprioceptive deficits in coronal plane, sagittal plane, and also with deficits in multiple planes. Parkhurst and Burnett (1994) concluded that impaired proprioception resulting from injury may degrade lumbar motion function and increase risk of reinjury. Restoration of low back proprioception after injury should be a focus during rehabilitation.

Gill and Callaghan (1998) assessed back proprioception in forty individuals using the Lumbar Motion Monitor (Chattecx Corp., Chattanooga, TN). Twenty participants with back pain and twenty participants with no pain were asked to reproduce a predetermined target position (20°of lumbar flexion) in standing and four-point kneeling positions. After a practice trial of 10 repetitions with visual feedback, each participant performed 10 times of position reproduction within 30 seconds. Participants were blindfolded for the test. A mean deviation from the target position was calculated for each subject. The group with LBP showed less accuracy than the healthy group revealing that differences in back proprioception do exist between individuals with and without back pain.

Newcomer et al. measured trunk repositioning error as a method of measuring back proprioception (Newcomer, Laskowski, Yu, Johnson et al., 2000). Twenty patients with LBP and twenty control subjects were tested in standing position, with their legs and pelvis partially immobilized. A 3Space Tracker system (Polhemus, Inc., Colchester, VT) was used to determine the motion and position of the trunk. Two electromagnetic sensors were placed on the skin over the T1 and S1 spinous process and secured with double-sided tape. The participants were tested in trunk flexion, extension, right-side bending, and left-side bending with eyes closed. Approximately 30%, 60%, and 90% of the maximum ROM in each movement direction were used as predetermined target positions. Repositioning error in patients with LBP was significantly higher than that of the control subjects in flexion positions, and significantly lower than that of the control subjects in extension positions. The authors believed that an increased repositioning error during flexion in the back pain group has clinical significance because many functional activities require trunk flexion. Trunk flexion is more complex than the movements in other directions because it requires coordination of trunk, hip, and pelvic muscle activations. Extension and lateral bending involve mainly abdominal muscles and erector spinae without the complex movements of the hips and pelvis. It was not clear why the repositioning error was decreased in extension, although the presence of pain may contribute to the awareness of body position.

Brumagne et al. also found that patients with LBP have a less refined position sense by measuring the repositioning error in sacral tilt (Brumagne et al., 2000). Twenty-three patients with LBP and twenty-one control subjects were tested in a sitting position with a piezoresistive electrogoniometer attached to the skin over the sacrum at spinous process of S2. Reproduction of predetermined sacral tilt angles were measured before, during, and after lumbar paraspinal muscle vibration. The results indicated that the repositioning accuracy in sacral position sense was significantly lower in the patient group before muscle vibration. Conversely, muscle vibration resulted in an undershooting of the target position in the control group and an improvement in position sense in the group with back pain. Based on these findings, Brumagne et al. concluded that altered paraspinal muscle spindle afference and altered central processing of lumbosacral position sense input may exist in patients with LBP.

Proprioception deficits in trunk flexion have been identified in non-athletes with back pain. The functional activities of these individuals mainly require trunk flexion. However, athletic activity usually requires athletes to perform accurate movements in all directions. Back proprioception deficits in athletes may be in more than one plane. The golf swing requires a golfer to bend and rotate trunk rapidly. Any deficits in back proprioception could alter the movement pattern and cause extra or abnormal stress to the lumbar spine. This could aggravate back injuries, resulting in vicious cycle. Evaluation of back proprioception in golfers with LBP was, therefore, needed in order to better understand their deficits and design appropriate rehabilitation programs.

4. Postural Stability in Low Back Pain

Interaction among the visual, vestibular, and sensorimotor systems is required for maintaining posture and balance (Luoto et al., 1998; Riemann, Myers, & Lephart, 2002). In addition, impairments in strength, coordination, or effective coupling of muscles in the lumbar and pelvis area can disturb postural stability (Luoto et al., 1998). Patients with LBP often have decreased trunk muscle strength, endurance, and mobility (Burton, Tillotson, & Troup, 1989; Mayer, Gatchel, Kishino et al., 1985; Roy, De Luca, & Casavant, 1989). This impairment in postural control was, therefore, suggested to be related to LBP (Luoto, Taimela, Hurri et al., 1996; Oddsson, Persson, Cresswell et al., 1999).

Nies and Sinnott found that individuals with LBP demonstrated significantly greater postural sway and were significantly less likely to be able to balance on one leg with their eyes closed (Nies & Sinnott, 1991). The sway velocity of COP in the patients with severe LBP was also observed to be significantly faster than that of the healthy group while standing on one leg with eyes open (Luoto et al., 1998). In the study of Mientjes and Frank (1999), subjects with and without LBP performed seven postural tasks while standing on both legs. These tasks involved manipulation of visual, vestibular, and proprioceptive input as well as body orientation. The RMS in the medial-lateral direction of the COP was significantly increased in the LBP group during tasks with eyes closed, especially when the complexity of the task was increased.

The significantly reduced postural stability in patients with LBP may suggest that central and peripheral balance control mechanisms were less effective in patients with LBP (Ebenbichler et al., 2001). It may also indicate the impairment of neuromuscular feedback loops at different levels of motor activation within the central nervous system (Ebenbichler et al., 2001). Good postural stability may not only contribute to a golfer's swing, but also reduce the chance of getting injured. Thus, it is important to assess the postural stability in golfers with LBP. The assessment of postural stability is also suitable to serve as an outcome measure when designing or conducting therapeutic exercises for injured golfers.

C. Methodology Considerations

1. 3D Biomechanical Analysis of Golf Swing

Linked segment models (LSMs) are often used for the mechanical analysis of human movement (Kingma, de Looze, Toussaint et al., 1996). This inverse dynamics approach models the full body or part of the body as a chain of rigid body segments interconnected by joints. Intersegmental reactive forces and moments are calculated using Newtonian mechanics for each body segment, starting at one end of the chain (Kingma et al., 1996). The assumptions for the inverse dynamics approach are: 1) each segment has a fixed mass which is located at its center of mass (COM); 2) each segment's COM remains fixed throughout the movement; 3) the connection between segments are considered as a hinge (or a ball and socket) joint; 4) each segment's mass moment of inertia about its COM (or about either proximal or distal joint) remains constant throughout the movement; 5) each segment's length remains constant throughout the movement (Winter, 1990). Three-dimensional LSM have been widely used for research of the lower extremity. Forces and moments of the knee joint are often calculated for various tasks, including walking, jumping, and landing. As for the spinal load, several 3D dynamic LSMs have been developed for lifting tasks using either top-down or bottom-up inverse dynamics approach (Kingma et al., 1996; Lavender & Andersson, 2000; Lavender, Li, Andersson et al., 1999; Plamondon, Gagnon, & Desjardins, 1994). They have proven to be valuable tools for evaluating spinal forces and moments.

Studies have tried to estimate spinal loads during the golf swing with the assistance of EMG (Hosea et al., 1990; Lim & Chow, 2000). Although the idea of using EMG-assisted model is reasonable, it is difficult to provide an accurate estimation for the trunk muscle forces during the golf swing because the relationship between EMG and muscle force is complicated. Surface EMG of the abdominal muscles is also difficult to record because of the influence of adipose tissue and signal artifact from the heartbeat. Although trunk EMG signals have been added into very complicated models for predicting spinal forces and moments in lifting studies (Cholewicki & McGill, 1994; Granata & Marras, 1993; Marras & Sommerich, 1991; McGill, 1992), it is important to realize that many factors influence the relationship between EMG and muscle force. These factors include the kinematics of the movement, the processing methods used, and the acquiring procedures (Redfern, 1992). In order for EMG to be used to assist in the prediction of muscle force, several conditions must be satisfied: 1) the muscle contraction must be in a static or controlled dynamic state; 2) the EMG-force relationship must be appropriate for the properties of the muscle and can be presented by a functional relationship or a model; 3) the given portion of the muscle must be sampled to prevent factors such as the length-tension relationship of the muscle confounding the EMG-force relationship (Marras, 1992). Only under these conditions can researchers make statements about the relative amount of muscle force during different work conditions or tasks (Marras, 1992). Most EMG-assisted models for spinal load estimation were developed for lifting tasks under slow or constant movement speeds. The golf swing, however, is a rapid movement involving a number of trunk muscles contracting at various speeds. As such, muscle length is also rapidly changing during the golf swing. These situations have a significant influence on the relationship between EMG and muscle force that affects the estimation of muscle forces.

EMG-assisted models and LSMs are tools for investigating the mechanisms of injury and the effects of technique during functional activities on the risk of injury. The complex EMG-assisted models provide insight as to how injury occurs with various levels of loads. On the other hand, the LSM can be a powerful tool for routine examination of physical demands of an activity (McGill, 2002). The choice of models should depend on the issues in question and need to be interpreted wisely in each case for their limitations and constraints (McGill, 2002). Because trunk muscle forces acting on the spine are not easily estimated through trunk EMG activities during the golf swing, it would be appropriate to use a dynamic LSM for the purpose of this study.

2. Assessment of Trunk and Hip Muscle Strength

Evaluations of trunk and hip muscle strength have been used to determine if there are differences between individuals with and without LBP. There are several methods for measuring muscle strength: isometric, isokinetic, and isoinertial. The selection of muscle strength testing should be based on the function of muscle performance during the tasks, such as maintaining the joint stability, posture, or the dynamics of motion. The modern golf swing restricts the hip turn to build torque in the back and shoulders during the backswing for maximum clubhead velocity at ball impact (Hosea & Gatt, 1996). Hip muscles play an important role in stabilizing the movement of the pelvis during the golf swing. Trunk muscles are responsible for rapid torso rotation, flexion, extension, and side bending in one combined movement. Thus, the measurement of isometric hip muscle strength and isokinetic trunk muscle strength would be appropriate for assessing the strength differences between golfers with and without LBP. Furthermore, it was suggested that rapid motion seems to be able to discern the loss of muscle function in patients with LBP better than slow motion (Ljunggren, 1993). Isokinetic test at slow speeds, however, are considered to reveal articular problems (Ljunggren, 1993). Evaluation of trunk strength at both fast and slow speed would provide more information regarding back problems in golfers with LBP as compared with their healthy counterparts.

3. <u>Assessment of Trunk and Hip Flexibility</u>

Low back pain is often associated with reduced flexibility of the trunk and hip muscles (Ellison, Rose, & Sahrmann, 1990; Mellin, 1988; Pope, Bevins, Wilder et al., 1985; Vad et al., 2004). In these studies, measures of the trunk and hip flexibility are usually performed to assess the ROM available to the spine and hip joint. Intra- and inter-rater reliability of the measurements of spinal and hip ROM have been considered to be acceptable or good (Alaranta, Hurri, Heliovaara et al., 1994; Ellison et al., 1990; Holm, Bolstad, Lutken et al., 2000; Keeley, Mayer, Cox et al., 1986; Klein, Snyder-Mackler, Roy et al., 1991; Pope et al., 1985; Vad et al., 2004). Therefore, the flexibility of the trunk and hip were assessed by measuring the ROM of spine and hip joint in this study.

4. Assessment of Back Proprioception

Methods for assessing conscious component of proprioception have been designed to measure joint position sense, kinesthesia, and the sense of tension. The joint position sense test is used more often in recent years for assessing back proprioception (Allison & Fukushima, 2003; Brumagne, Lysens, & Spaepen, 1999; Fujiwara, Miyaguchi, Toyama et al., 1999; Gill & Callaghan, 1998; Koumantakis, Winstanley, & Oldham, 2002; Lam et al., 1999; Maffey-Ward, Jull, & Wellington, 1996; Newcomer, Laskowski, Yu, Johnson et al., 2000; Preuss, Grenier, & McGill, 2003; Swinkels & Dolan, 1998). It measures the accuracy of position replication by calculating repositioning error. The joint position sense test can be tested actively or passively in both open and closed kinetic chain positions (Riemann et al., 2002).

Researchers have suggested that assessing proprioception deficits actively during normal self-paced movements may be more functionally relevant (Brumagne et al., 2000; Gill & Callaghan, 1998; Koumantakis et al., 2002; Newcomer, Laskowski, Yu, Johnson et al., 2000; Swinkels & Dolan, 2000). Accuracy of lumbar spine repositioning was found to be better in the standing position compared to the sitting and four-point kneeling position (Preuss et al., 2003). Proprioception deficits are more likely to represent abnormal or insufficient afferent information transferring from mechanoreceptors when tested in midrange active movements. It is because these receptors provide the majority of proprioceptive information regarding position and movement of peripheral joints under these conditions (Brumagne, Lysens, Swinnen et al., 1999; Gandevia, McCloskey, & Burke, 1992; Koumantakis et al., 2002). It was found that the accuracy of the spinal position sense in the outer range (80% of available range) of movement is better than that in the inner range (20% of available range) (Allison & Fukushima, 2003). Six trials of testing may be required to derive a representative value of accuracy and precision in spinal reposition (Allison & Fukushima, 2003). Furthermore, the mobility of the pelvis and legs was suggested to decrease while measuring the back repositioning errors (Newcomer, Laskowski, Yu, Johnson et al., 2000). Although immobilization of the pelvis and legs is not able to simulate functional activities very well, unrestricted position may neutralize the differences between the patient and the healthy groups by providing additional afferent input from the lower extremities and pelvis (Newcomer, Laskowski, Yu, Johnson et al., 2000).

5. Assessment of Postural Stability

The most commonly used method for static postural stability assessment is single-leg standing balance test. It requires the center of gravity of the body to be maintained within a narrow and short base of support. Thus, the importance of the segmental control in the frontal plane is increased (Riemann et al., 2002). Laboratory measurements of postural stability usually involve standing on a force plate with eyes open and eyes closed. Force plate measures the distribution of the applied forces that can be calculated as COP or the variability of forces in horizontal or vertical plane to represent postural stability. Variables that have been used for comparing the differences in postural stability between individuals with and without LBP include COP shift and RMS of the COP sway in anterior-posterior and medial-lateral direction, and the sway velocity of the COP (Luoto et al., 1998; Mientjes & Frank, 1999; Nies & Sinnott, 1991). Among these evaluated variables, RMS of the medial-lateral sway of the COP and the sway velocity of the COP have been found to be significantly increased in the individuals with LBP. They can be used as indicators when evaluating single leg standing balance. It was also reported that the sway velocity of the COP during the single-leg standing balance test was the most sensitive parameter for evaluating postural stability (Luoto et al., 1998).

D. Summary

The golf swing requires the upper body to rotate on the hips and pelvis during the backswing, and then uncoil forcefully for ball impact and follow through. This rapid and powerful movement can produce a tremendous amount of force and torque that may lead to injury of the lumbar spine and back muscles. With more and more players of all ages and fitness levels participate in this sport, the incidence of back disorders has increased (Hosea & Gatt, 1996). Both professional and amateur golfers could suffer from low back injuries (Gosheger et al., 2003; Hosea & Gatt, 1996; McCarroll & Gioe, 1982).

Identifying causes of low back injuries in golfers is not easy, as the injuries may result from a combination of factors, such as the forward and backward twisting motion of the upper and lower torso, poor swing posture, and insufficient physical conditions. Thus, in addition to acquiring a biomechanically efficient golf swing, golfers are suggested to work on their muscle strength, flexibility, balance, and coordination for injury prevention and rehabilitation. However, scientific studies are not enough to support an appropriate training or injury prevention program specifically designed for protecting golfer's back. Training without knowing on what to focus could result in a strength imbalance that may lead to further injury. Investigating the physical and biomechanical differences between golfers with and without LBP is imperative for identifying the potential injury risk factors among golfers. This information can serve as the basis of injury prevention and rehabilitation programs. It will then contribute to optimal performance without pain.

III. METHODOLOGY

A. Experimental Design

This comparison study evaluated and compared two groups of golfers – with and without LBP. The independent variable was group and the dependent variables were the trunk kinematics, lumbar spinal kinetics, trunk and hip strength, trunk and hip flexibility, back proprioception, and single-leg standing balance. Specifically, the variables that were evaluated in each specific aim are listed below. Additionally, spinal kinetics at 7 points of the golf swing (Table 3.1) and individual trunk and hip muscle strength were also evaluated for both groups of golfers.

Specific aim 1 – The trunk kinematics during the golf swing, including maximum X factor normalized by the maximum trunk rotation angle toward the non-lead side in neutral position, maximum crunch factor, maximum incline factor, trunk extension angle at the end of swing, and maximum spinal rotation velocity. Trunk movements were calculated relative to the pelvic movements.

Specific aim 2 – The lumbar spinal kinetics during the golf swing, including maximum forces in three anatomical axes (compression force, anterior-posterior shear force, and lateral shear force) and maximum moments about the three anatomical axes (flexion-extension moment, lateral bending moment and vertical rotation moment) at the L5/S1 joint.

Specific aim 3 – Strength ratios of trunk extension/flexion and right/left rotation. Side-to-side strength difference of each hip muscle group, including hip flexors, extensors, abductors, and adductors.

Specific aim 4 – Trunk flexibility, including trunk extension, flexion, and rotation. Hip flexibility, including bilateral hip flexion, extension, abduction, adduction, internal rotation, external rotation, FABERE's distance, and hamstring flexibility.

Specific aim 5 – Back proprioception, including active spinal repositioning errors in the sagittal plane (flexion and extension), frontal plane (right and left side bending), and horizontal plane (right and left rotation).

Specific aim 6 – Single leg standing balance, including sway velocity of the COP while standing on one leg with eyes open or eyes closed.

Table 3.1 Description of 7 analyzed swing points

Address	Point where club begins to move
Take-Away	Point where club shaft is parallel to the ground during backswing
Top	Point where club begins to be pulled down
Acceleration	2/3 of the time from top to impact
Impact	Point where clubhead contacts the ball
Follow through	Point where club shaft is parallel to the ground after impact
Finish	Point where the clubhead stops the swing

B. Subject Characteristics

Golfers with and without LBP were recruited in this study. The use of previous literature and a conservative estimate determined 16 pairs of subjects for the study groups based on an alpha level of 0.01 and a power of 0.827. The variable selected for power analysis was the maximum X factor normalized by the maximum trunk rotation angle toward the non-lead side in neutral position. The average values of this variable in the LBP and pain free groups were 108.3 ± 20.0 % and 88.0 ± 24.9 %, respectively, in the study of Lindsay and Horton (2002). Subjects were invited to participate through the use of posted flyers in local golf courses and rehabilitative clinics (Appendix A). Subject with previous LBP filled out a modified Oswestry questionnaire (Fritz & Irrgang, 2001) (Appendix B), pain scale, and pain diagram (Appendix C). Subjects with previous LBP underwent a basic neurological examination to screen for nerve root compromise (Appendix D). All subjects needed to meet the following criteria:

Inclusion Criteria

- General:
    1. USGA handicap < 20
    2. Age: 18-65 years
- Back Pain group:
    1. Subjects had symptoms of mechanical LBP within two years prior to testing resulted in time lost from golf participation
    2. Subjects had recurrent mechanical LBP within two years prior to testing
    3. The worst episode of LBP within two years prior to testing had a modified Oswestry questionnaire score greater than 24 and required physical treatment
    4. LBP resulted from golf or was aggravated by golf

5. LBP was localized over right or central lumbosacral area for the right-handed golfer and left or central lumbosacral area for the left-handed golfer
6. Asympatomatic during the time of testing

- Healthy group:
  1. Subjects must not have LBP within two years prior to testing
  2. Age, gender, and golf handicap were matched with the back pain group

Exclusion Criteria

- Subjects with a history of previous back surgery, vertebral compression fracture, nerve root compromise, neurologic deficits, current lower extremity symptoms, current lumbar radiculopathy or a history of the condition, and symptoms of vertigo or dizziness were excluded.

C. Instrumentation

1. Peak Motus 3D Video Motion Analysis System

Kinematic data of the golf swing was collected using the Peak Motus System v.8.2 (Peak Performance Technologies, Inc., Englewood, CO). It is a 3D analysis system with eight optical cameras (120 Hz) (Pulnix Industrial Product Division, Sunnyvale, CA) that were placed at a distance of 4 m around two force plates. The capture volume was $3.0 \times 4.3 \times 2.9 \text{ m}^3$. Calibration was done using the wand calibration method (wand length = 0.914 m) according to the manufacturer's guidelines. A root mean square error of 0.002 meters and 0.254 degrees was obtained in the Neuromuscular Research Laboratory for determining the measurement accuracy of position and angular data. Anthropometric measurements, reflective markers with a diameter of 0.025 m, and coordinate data collected from the camera recordings allowed for calculations of the center of rotation (ankle, knee, hip, trunk, shoulder, elbow, and wrist) and the segmentally embedded coordinate systems as described by Vaughan et al. (Vaughan, Davis, & O'Connor, 1991).

2. Kistler Force Plate

Ground reaction forces were obtained with two force plates (Kistler Instrument Corporation, Amherst, NY). The force plates were placed 14 cm apart and connected directly to the Peak Motus System to determine ground reaction forces. The coordinates of force plates were calibrated in accordance with the global reference frame and registered in Peak Motus software (Version 8.2). Ground reaction force data were collected at 1,200 Hz during the golf swing testing and at 100 Hz during the balance testing.

3. Biodex Isokinetic Dynamometer

Trunk and hip muscle strength was assessed with the Biodex System III Multi-Joint testing and Rehabilitation System (Biodex Medical Inc., Shirley, NY). Torque values were automatically adjusted for gravity by the Biodex Advantage Software v.3.2 (Biodex Medical Inc., Shirley, NY). Calibration of the Biodex dynamometer was performed according to the specifications outlined in the manufacturer's service manual. The trial-to-trial and day-to-day reliability and validity of torque measurement of the

Biodex System III were all previously established with intraclass correlation coefficients (ICC) reported to be 0.99-1 (Drouin, Valovich-mcLeod, Shultz et al., 2004).

4. <u>The MotionMonitor 3D Motion Capture System</u>

Back proprioception and trunk flexibility were assessed with the MotionMonitor (Innovative Sports Training, Chicago, IL) 6 degrees of freedom electromagnetic motion analysis system. The MotionMonitor is a computer interfaced with an electromagnetic transmitter (direct current) mounted on a wooden base. The mounted transmitter emitted an electromagnetic field with a 12 feet radius effective in all directions. Electromagnetic sensors were interfaced with the computer and relayed information concerning position and orientation (X, Y, Z coordinates as well as yaw, pitch and roll) where within the electromagnetic field. The data collection rate for the current study was 100 Hz. The sensors were stabilized and secured to the subject's skin using 3M double-stick discs or surgical tape. Any motion by the subject that occurred within the electromagnetic field was relayed back to the computer, creating a 3 dimensional (3D) computer-generated humanoid representation. A root mean square of 0.004 meters and 0.3 degrees was obtained in the Neuromuscular Research Laboratory for determining the accuracy of the receivers in measuring the position and orientation.

5. <u>Standard Goniometer</u>

Hip flexibility or range of motion (ROM) was be measured using a standard goniometer. A small level was attached parallel to the stationary arm of the goniometer to verify correct orientation.

D. Testing Procedures

1. Subject Preparation

All subjects provided written informed consent approved by the University of Pittsburgh Institutional Review Board prior to participation. Subjects with previous LBP within the past two years completed a modified Oswestry questionnaire, pain scale, and pain diagram. All testing procedures were taken place at the Neuromuscular Research Laboratory and the Golf Fitness Laboratory in the Center for Sports Medicine, University of Pittsburgh Medical Center.

2. Order of Testing

The order of testing was arranged as the following sequence: 1. back proprioception assessment, 2. trunk rotation strength assessment, 3. trunk kinematic and kinetic analysis during the golf swing, 4. trunk flexion/extension strength assessment, 5. trunk and hip flexibility assessment, 6. postural stability assessment, and 7. all hip strength assessments. The maximum effort required to perform trunk and hip strength assessments might affect a subject's performance in assessment of back proprioception and balance, therefore back proprioception and balance were ordered as such. This order might also provide investigators to have enough time for preparing the equipment of each subsequent testing.

3. Back Proprioception Assessment

Spinal repositioning error was measured as a method of assessing back proprioception. Spinal repositioning error was measured using the MotionMonitor 3D motion capture system at the frequency of 100Hz. Subjects stood with their feet shoulder-width apart, eyes closed, and arms crossed at their chest. To decrease movement of the pelvis, the subjects were partially immobilized with a custom pelvis stabilizing apparatus (Figure 1). The movement of the pelvis was restricted by two bars, one placed in front and the other behind during trunk flexion, extension and rotation. The bar in front was placed just below the anterior superior iliac spine and the bar behind was placed below the posterior superior iliac spine. For the trunk lateral bending testing, the subjects turned 90 degrees from the previous standing position. The two bars were placed just below the right and left iliac crest. The force of the fixation were adjusted without causing discomfort to the subject.

Two sensors were attached to the subject's skin at the first segments of the thoracic and sacral spine (T1 and S1) using 3M double-sided tape and surgical tape (3M Health Care, St. Paul, MN) for measuring the spinal position (Figure 3.1). An experienced physical therapist manually palpated the spine (Chaitow, 1997) to determine the spinal level, used by the MotionMonitor software to create a 3D image of each subject. General palpation was recommended by the MotionMonitor manual for proper setup. The subjects were asked to perform maximal trunk flexion, extension, and lateral bending and rotation to both the left and right side. Total pain free ROM for trunk flexion, extension, lateral bending and rotation were assessed prior to testing procedures using the MotionMonitor system. Subjects were then blindfolded to eliminate visual input. Subjects were placed in a target position that is approximately 80% of the total pain free ROM for a specific direction for 4 seconds (i.e. flexion, extension, lateral bending, or rotation). A biofeedback sound from the MotionMonitor system was provided as an audible cue for helping the subjects to maintain this test position. The subject then returned to the neutral position and attempted to reproduce the target position without any assistance. Six trials were taken for each of the four directions of trunk movement. The order of testing trunk position was randomly assigned to each subject. The mean absolute difference in degrees was calculated for each position. The ICC and SEM for the protocol outlined above was determined by the primary investigator of this study to be 0.16 and 0.97º for trunk flexion, 0.37 and 0.67º for trunk extension, 0.06 and 0.61º for trunk right rotation, 0.50 and 0.50º for trunk left rotation, 0.23 and 0.60º for trunk right side bending, and 0.46 and 0.15º for trunk left side bending.

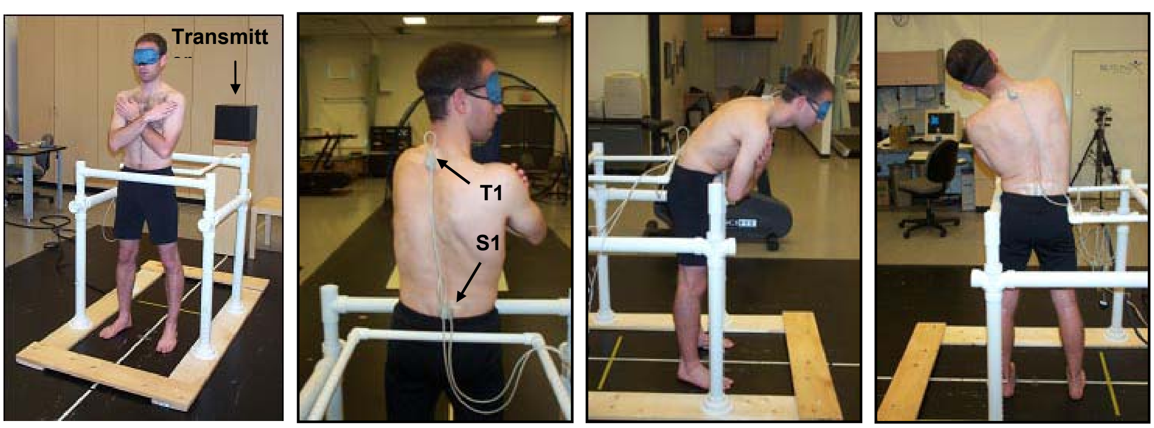


Figure 3.1 Back Proprioception assessment

4. <u>Kinematic and Kinetic Analysis of Trunk during the Golf Swing</u>

Anthropometric measurements of the lower extremity were taken including body mass and height, anterior-superior iliac spine (ASIS) breadth, thigh, calf and foot length, mid-thigh and calf circumference, knee diameter, malleolus height, malleolus width, and foot breadth. Anthropometric measurements of the upper extremity included upper arm length, forearm length, forearm diameter, hand length, hand diameter, and hand width. Subjects were fitted with reflective markers (0.025 m diameter) at the following lower extremity landmarks: the posterior heel, lateral malleolus, second metatarsal head, femoral epicondyle, ASIS, and sacrum. Reflective markers will also be placed at the following upper extremity landmarks: acromion, lateral epicondyle of the humerus, wrist, and T4 level of the spine. Eight markers were attached to wands (distance of 0.09 m from the skin) and secured with Velcro straps on the lateral side of the mid-thigh, mid-calf, mid-forearm, and mid-upper arm. Two markers were placed on each side of the body at the L5/S1 level to locate the center of lumbo-sacral joint. Three markers were used to define the upper portion of the lumbar spine - right and left ribs and the spinal process at the T12/L1 level of the spine (Morgan, Cook, & Banks, 1999). Two markers were placed on the golf club to identify the phases of the golf swing (Figure 3.2).

A self directed warm-up, stretching, and practice shots were provided prior to data collection. A static calibration was collected for each subject prior to the testing as well. Subjects were instructed to stand in anatomical position with their feet shoulder-width apart in the capture volume. The joint angles calculated during the golf swing were normalized by subtracting the joint angles calculated from the static trial during data reduction. Subjects hit golf balls with their own driver to represent the actual swing pattern experienced while playing. Subjects stood with one foot on each force plate during the golf swing. Subjects hit 10 shots off an artificial turf mat into a screen approximately five meters away.

Figure 3.2 3D biomechanical analysis of golf swing

5. <u>Trunk and Hip Strength Assessment</u>

Trunk Strength

For torso rotation testing, subjects were seated in an upright position (Figure 3.3a). The rotational axis of the torso rotation attachment was aligned with the long axis of the spine of each subject. The lumbar pads were adjusted to firmly secure the subjects. The input assembly was lowered so that the chest pad could contact the subjects' chests approximately two inches below the level of the clavicles. The subjects were secured with back and thigh stabilization straps to minimize extraneous body movements and momentum. Thigh pads were aligned approximately 3" proximal to the medial femoral epicondyles of each leg. Subjects performed left and right torso rotation for 5 repetitions at 60º/sec and 10 repetitions at 120º/sec. There was a one-minute rest between the two speeds of testing. The reliability of trunk rotation isokinetic strength testing had been previously established in the Neuromuscular Research Laboratory with intraclass correlation coefficients (ICC) to be 0.73-0.85 for the peak torque/body weight (PT/BW) at the speed of 60º/sec and 120º/sec.

For torso flexion and extension testing, subjects were seated in a semi-standing position with the seat in down position approximately 15º and 15º flexion in knee joint (Figure 3.3b). The axis of rotation of the dynamometer resistance adapter was aligned with the ASIS. Torso straps, the clavicle pads on the torso straps, and lumbar pad were adjusted and applied firmly for maximum patient restraint and comfort. The subjects' pelvis and thighs were stabilized with straps designed to minimize extraneous body movements and momentum. Subjects performed torso flexion and extension for 5 repetitions at 60º/sec and 10 repetitions at 120º/sec with a one-minute rest between the testing speeds. The reliability of trunk flexion and extension isokinetic strength testing had been previously established with intraclass correlation coefficients (ICC) reported to be 0.79-0.92 for the PT/BW of trunk flexion and 0.74-0.90 for the PT/BW of trunk extension at the speed of 60º/sec and 120º/sec (Delitto, Rose, Crandell et al., 1991). Karatas et al. also reported reliability for this testing with a 0.89-0.95 ICC for the peak torque of trunk flexion and 0.80-0.92 ICC for the peak torque of trunk extension (Karatas, Gogus, & Meray, 2002) at the speed of 60º/sec and 90º/sec.

a. rotation

b. flexion/extension

Figure 3.3 Torso strength assessment

Hip Strength

Subjects were asked to perform isometric contraction of hip abduction, adduction, flexion, and extension with the greater trochanter aligned with the axis of rotation of the dynamometer resistance adapter (Figure 3.4a, b). Subjects were tested in side-lying position with hip joint in neutral position during the testing of hip abduction (Perrin, 1993). During the testing of hip adduction, subjects were placed at 20° of hip abduction in side-lying position. During the testing of hip flexion and extension, subjects were placed at 15° of hip flexion in supine position (Perrin, 1993). Subjects were also secured using torso and pelvic straps in order to minimize extraneous body movements and momentum. Each subject performed three isometric contractions in each direction on both legs. Each isometric contraction lasted for 5 seconds. A 10-second-rest interval was provided between contractions.

Practice trials were provided for each muscle strength testing to ensure patient understanding and familiarity. Practice included three sub-maximal contractions followed by three maximal contractions. After one minute rest followed by the practice trials, each subject was instructed to perform a maximal effort with each contraction during the testing.

a. abduction/adduction

b. flexion/extension

Figure 3.4 Hip strength assessment

6. Trunk and Hip Flexibility Assessment

Trunk Flexibility

Trunk flexibility was derived from the maximum ROM of trunk in each movement direction measured during the back proprioception assessment. Prior to the test of trunk position reproduction, maximum ROM of trunk flexion, extension, and rotation were measured using the Motion Monitor system for deciding the trunk-repositioning target. The maximum angles of the trunk movements served as the flexibility of the trunk as well. During these measurements, each subject's pelvis was stabilized with the stabilization device as described in the session of back proprioception assessment (Figure 3.1).

Hip Flexibility

Hip joint flexibility was measured passively based on the methods described in the textbook of Norkin and White (Norkin & White, 1995). Hip flexion was measured with the subjects in a supine position with the hip in 0º of abduction, adduction, and rotation. Both knees were placed in full extension in the beginning of the test. The knee of the test leg was moved into flexion as the hip was moved into flexion. The pelvis was stabilized to prevent rotation and tilting. Hip extension was measured with the subjects in a prone position with both legs straight and pelvis stabilized.

Hip abduction and adduction were measured with the subjects in a supine position with hip in 0º of flexion, extension, and rotation. The pelvis was stabilized to prevent rotation and lateral tilting. During the measurement of adduction, the contralateral hip was placed abducted to avoid contact with the test leg.

Hip internal and external rotation were measured with the subjects sitting on the measurement table with hips and knees flexed to 90º. During the measurement, hips were kept in 0º of abduction and adduction. The femur of the testing leg was stabilized to prevent extra hip movements. Lateral pelvic tilt was avoided.

FABERE's test was performed with the subjects in a supine position. FABERE's distance was measured in centimeters from the knee (lateral epicondyle of the femur) to the horizontal with the hip in flexion, abduction, and external rotation while the ipsilateral ankle rested on the contralateral knee.

Hamstring flexibility was measured in a supine position using the active knee extension test. The hip was passively flexed until the thigh is vertical. This thigh position was maintained throughout the test while the opposite leg was fully extended.

The foot of the leg being tested was kept relaxed while the leg was actively straightened until the thigh began to move from the vertical position. The minimum angle of knee flexion with the thigh in the vertical position was measured.

The intra-rater reliability of goniometric measurement of hip ROM had been previously established with ICC reported to be 0.82 for flexion, 0.94 for extension, 0.86 for abduction, 0.5 for adduction, 0.9 for external rotation, and 0.9 for internal rotation (Holm et al., 2000). Gajdosik also reported reliability of hamstring flexibility measurement to be 0.99 ICC (Gajdosik & Lusin, 1983)

7. Postural Stability Assessment

Postural stability was assessed using a Kistler force plate (Kistler Corporation, Amherst, NY) at the frequency of 100Hz. Each subject was asked to complete a single-leg standing balance test for each leg under two conditions (eyes open and eyes closed) (Figure 3). Three ten second trials were collected for each leg under each condition. Prior to testing, the subject was asked to remove shoes and socks. During the testing session, the subjects were instructed to remain as erect as possible with feet shoulder width apart and hands on hips. Subjects were instructed to focus on a target located approximately 2 meters in front of them at eye level during the testing session with eyes open. During the testing session with eyes closed, the subjects were instructed to focus on the target for balance first then close their eyes for data collection. Luoto et al. reported that the test-retest measurements of the sway velocity of the COP were within 95% deviation during the single-leg standing balance test (Luoto et al., 1998). Thus, the reliability of the measurement was acceptable.

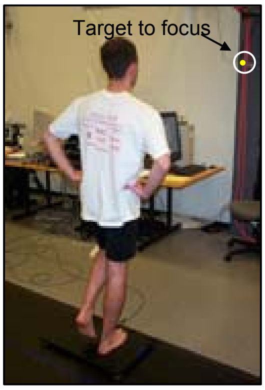


Figure 3.5 Single-leg standing balance test

E. Data Analysis

1. Data Reduction

Golf Swing Analysis

Kinematic data of the golf swing were filtered using an optimized cutoff frequency (Jackson, 1979). The X factor was calculated using the shoulder rotation angle subtracted by the pelvic rotational angle. Shoulder and pelvic rotation angle were calculated based on the orientation of shoulder markers and the orientation of ASIS markers with respect to the medial-lateral axis of the global coordinate system in the horizontal plane. Spinal rotation velocity was the change of the X factor over a discrete period of time. Trunk anterior/posterior tilt angle was calculated as the angle of the torso segment (middle of shoulder markers to middle of the markers at L5/S1 level) with respect to the pelvic anterior/posterior tilt in the anatomical plane. Trunk lateral bending angle was calculated as the angle between the torso segment and the ASIS segment, subtracted by 90 degrees. Lumbar lateral bending angle was calculated as the angle between the lumbar segment (middle of the markers on the side of the ribs at T12/L1 spinal level to middle of the markers at L5/S1 level) and the ASIS segment, subtracted by 90 degrees. The joint angles calculated during the golf swing were normalized by subtracting the joint angles calculated from the static trial. Maximum X factor was then normalized to the maximum trunk right rotation angle measured in the back proprioception assessment. Spinal rotation velocity, trunk anterior/posterior tilt, and trunk and lumbar lateral bending angles were derived for the calculation of crunch factor (lumbar lateral bending angle multiplied by spinal rotation velocity) and incline factor (trunk anterior/posterior tilt angle multiplied by trunk lateral bending angle).

Raw analog data from the two force plates was used to calculate ground reaction forces that were used for the calculation of spinal loads using an inverse dynamics procedure. Spinal forces were normalized to the subject's body weight. Spinal moments were normalized to the product of the subject's body weight and height.

AboutGolf's Sim Sensor and software (About Golf Limited, Maumee, Ohio) were used to determine the best 5 shots of each subject based on the estimated driving distance. Kinematic and kinetic data of each subject's best 5 shots were averaged for data analysis.

- *Dynamic 3D LSM for the Spinal Loads*

In this study, three dimensional spinal forces and moments at L5/S1 joint during the golf swing were computed using a bottom-up dynamic 3D LSM. L5/S1 joint was chosen because most of the lumbar spinal joint degeneration and disc herniations in golfers occur at the L3/L4, L4/L5, or L5/S1 level, and the mechanical load can be expected to be the highest at the lowest intervertebral disc (Hosea & Gatt, 1996; Kingma et al., 1996; Sugaya et al., 1999). A bottom-up dynamic 3D LSM was used because the speed (120 Hz) of the optical cameras that were used in this study was not fast enough to catch markers on the wrists, hands, and club during the downswing with a normal swing speed. This results in incapability of using a top-down dynamic 3D LSM to perform accurate calculation for the spinal loads. A pilot study done by the principal investigator of this study had validated a bottom-up dynamic 3D LSM with a top-down model by calculating the spinal forces and moments at lumbo-sacral joint during the slow motion of golf swing. The models and the calculations were similar to the lifting studies described by Kingma et al. and Kuo et al. (Kingma et al., 1996; Kuo, Chen, Wei et al., 1996). One segment for the golf club was added to the top-down model. The validation revealed reasonable to good agreement between the two kinetic analyses (Table 3.2). The coefficient of multiple correlation also showed good reproducibility of the spinal load calculations within each golfer's swings using the bottom-up model (Table 3.3). Thus, a bottom-up dynamic 3D LSM was adopted in an attempt to estimate the spinal loads during the golf swing in this study. Another advantage of using a bottom-up dynamic 3D LSM was that the trunk was excluded from the calculation (Kingma et al., 1996). Trunk tissues contain a relatively wide range of densities that makes it difficult to obtain a reliable estimate of the trunk center of mass (COM). The trunk COM is continuously moving due to breathing, and the trunk is not a rigid body. A LSM does not compensate for movements of the COM within body segments because the segments are supposed to be rigid. Additionally, the trunk is the segment with the largest mass that could cause strong effects of errors in the determination of the COM on the lateral bending and flexion-extension torque (Kingma et al., 1996).

Table 3.2 Validation of the LSM for golf swing

Coefficients of correlation and RMS errors of difference between the time series of the bottom-up and top-down calculated force and moment at the L5/S1 joint. Median values and ranges over 13 trials of 6 golfers slow swings are presented.

	Coefficients of correlation			RMS differences	
	Median	Range		Median	Range
Forces			**Forces (N)**		
Anterior-posterior	0.940	0.557 - 0.992	Anterior-posterior	17.4	3.1 - 30.5
Right-left	0.983	0.763 - 0.995	Right-left	11.6	4.8 - 25.7
Compressive	0.938	0.708 - 0.987	Compressive	16.6	9.1 - 35.3
Moments	0.825	0.609 - 0.942	**Moments (Nm)**		
Lateral bending	0.973	0.936 - 0.997	Lateral bending	11.1	6.4 - 21.6
Flexion-extension	0.914	0.645 - 0.968	Flexion-extension	13.2	4.9 - 17.3
Rotation			Rotation	9.5	5.8 - 16.2

RMS: root-mean-square

Table 3.3 Reproducibility of the bottom-up 3D dynamic LSM for golf swing

The coefficients of multiple correlation for the reproducibility of spinal loads calculation. The bottom-up 3D dynamic LSM was used. Values were derived from 3 golf swings of each subject. Ranges over 5 golfers are presented.

	Forces		**Moments**
Anterior-posterior	0.82 - 0.95	Lateral bending	0.96 - 0.98
Right-left	0.88 - 0.98	Flexion-extension	0.94 - 0.99
Compressive	0.91 - 0.98	Rotation	0.92 - 0.97

3D: three dimensional, LSM: linked segment model

- *The Calculation Procedure of Spinal Loads*

A customized program, using Matlab Version 6.0 Release 12 (The Mathworks, Inc., Natick, MA), was used for the calculation of spinal forces and moments. The steps of calculation process are shown in Figure 3.6. Three-dimensional coordinates of markers and joint centers, ground reaction forces, and subject's body mass was derived from the Peak Motus system and input to the Matlab program. Coordinates of joint centers were calculated within the Peak Motus software based on the methods described by Vaughan et al. (Vaughan et al., 1991). Anthropometric data, including segment length, weight (Webb Associates, 1978), moment of inertia (Chandler, Clauser, McConville et al., 1975; Webb Associates, 1978), and center of mass (Dempster, 1955) were calculated first. The references used for the calculation of anthropometric data can also be found in the book of Occupational Biomechanics (Chaffin, Andersson, & Martin, 1999). After the calculations of linear and angular velocity and acceleration of each body segment, intersegmental forces and moments in three-dimensions were calculated based on the inverse dynamic analysis procedure (Kingma et al., 1996; Kuo et al., 1996). According to the assumptions of the inverse dynamics approach, every body segment is considered as a "free body". Adjacent segments are connected to each other at the joint center. The analysis starts from one end of the chain of rigid segments. Figure 3.7 shows such a body segment and the forces and moments applying at it (Kingma et al., 1996). This is an example of a calculation in a two-dimensional plane. At each instant of time, the body segment was subject to the following equations of motion (Kingma et al., 1996):

$$\sum_{k=1}^{p} F_k + mg = ma$$

$$\sum_{k=1}^{p} ((\nu_{r,k} - \nu_{com}) \times F_k + \sum_{l=1}^{q} M_l = d(I\omega)/dt = (d(I)/dt)\omega + I\alpha$$

F_k : all p external and intersegmental forces k, applied at the body segment
m: segment mass; g: gravity vector; a: segment linear acceleration
ω: segment angular velocity; α: segment angular acceleration
I: moment of inertia
$\nu_{r,,k}$: point of application of force k
ν_{com} : segment center of mass
x: vector product
M_l: all q moments l, applied at the body segment

The bottom-up dynamic 3D linked segment model used in this study consisted of seven segments – bilateral feet, lower legs, thighs and a pelvis. Calculation of the spinal loads started from the ground reaction forces, followed by the forces and moments in the ankle, knee and hip joints, and ended at the lumbo-sacral joint. Forces and moments calculated in the global coordinate system were transformed into the local coordinate system of each body segment using a transformation matrix in order to represent the actual loads applied in each connected joint and the lumbo-sacral joint during the golf swing.

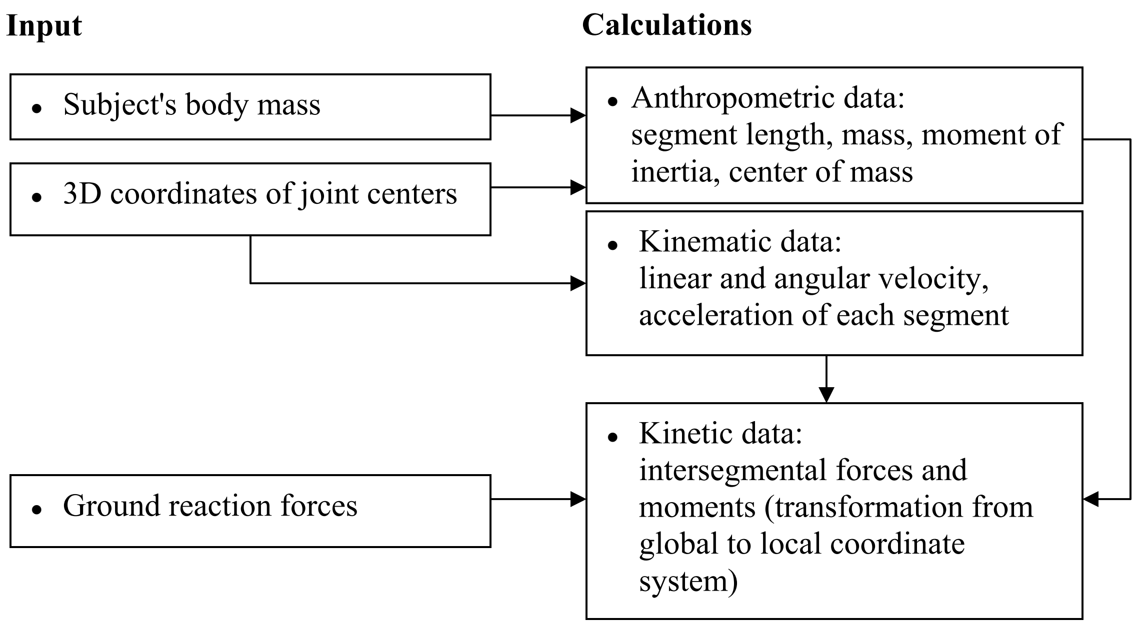


Figure 3.6 A schematic presentation of the input and calculations of the 3D dynamic LSM used in MATLAB program

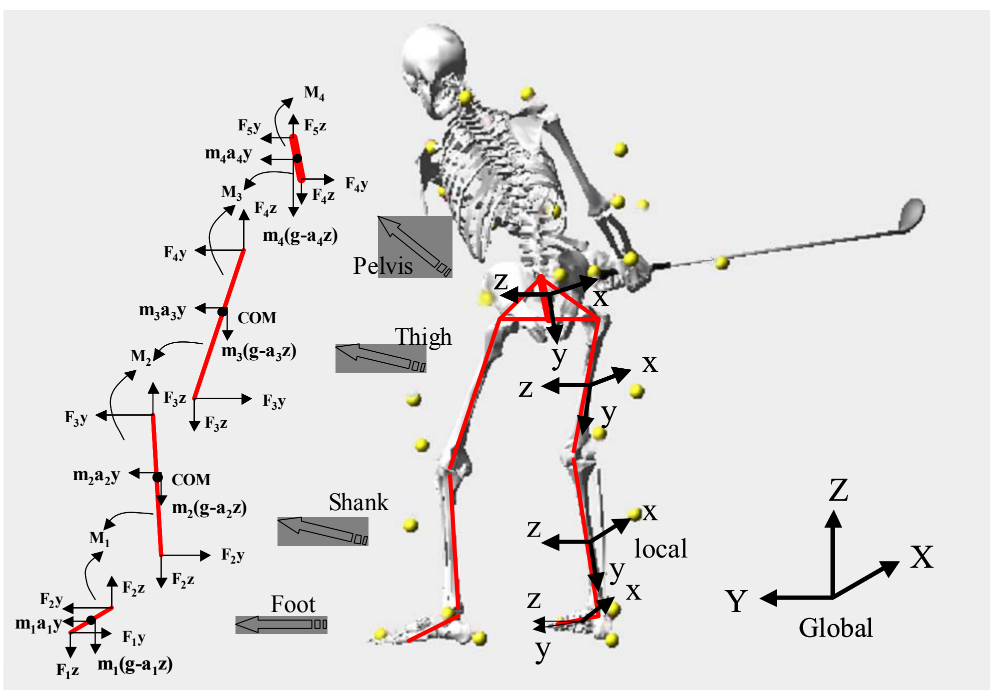


Figure 3.7 A 2D example of free-body diagrams during the inverse dynamic calculations. 3D global and local coordinates of each segment are also shown

Note: COM – center of mass, F – force, M – moment, m – mass, g – gravity, a – acceleration.

Trunk and Hip Strength Assessment

The peak torque/body weight ratio of each muscle strength measurement was derived from the Biodex for data analysis and compared between groups. It was used to calculate trunk strength ratios and side-to-side strength differences in each hip muscle group. Trunk strength ratios included trunk extension/flexion and right rotation/left rotation. Side-to-side strength difference of each hip muscle group was calculated as: absolute value of [(right hip strength – left hip strength)/maximum strength of both sides] x 100%.

Back Proprioception Assessment

Back proprioceptive data were filtered using a dual-pass fourth-order Butterworth filter with a cut-off frequency of 10 Hz within the Motion Monitor software. A customized program using Visual Basic for Applications (Microsoft Inc, Redmond WA) was used to quantify the degree error during active trunk position reproduction in each movement direction.

Postural Stability Assessment

Sway of the COP in centimeters while standing on one leg with eyes open and eyes closed was derived from the force plate data. The sway velocity of the COP (cm/sec) was also calculated using total sway distance divided by testing time.

2. <u>Statistical Analysis</u>

Descriptive statistics were analyzed to assess means and standard deviations between the back pain and healthy groups. SPSS 11.0 (SPSS, Chicago, IL) and BMDP statistical software (BMDP statistical software, Inc. Saugus, MA) were used for data analysis.

Trunk Kinematics During the Golf Swing

A one-tailed dependent t-test was used to determine significant differences in the maximum X factor and the maximum X factor normalized by maximum trunk right rotation angle in neutral position (Hypothesis 1.1) between golfers with and without LBP. Statistical significance was considered at the $p<0.05$ levels.

One-tailed dependent t-tests were used to determine significant differences in the maximum crunch factor, maximum incline factor, and the angle of trunk extension at the end of swing between golfers with and without LBP (Hypothesis 1.2). Statistical significances were considered at the $p<0.0167$ after Bonferroni's correction.

A one-tailed dependent t-test was also used to determine significant differences in the maximum spinal rotation velocity between golfers with and without LBP (Hypothesis 1.3). Statistical significance was considered at the $p<0.05$ levels.

Lumbar Spinal Kinetics During the Golf Swing

One-tailed dependent t-tests were used to determine significant differences in the maximum spinal forces (Hypothesis 2.1) and the spinal forces at 7 swing points, including compression force, anterior-posterior shear force, and lateral shear force at L5/S1 level, between golfers with and without LBP. Statistical significances were considered at the $p<0.0167$ after Bonferroni's correction.

One-tailed dependent t-tests were used to determine significant differences in the maximum spinal moments (Hypothesis 2.2) and the spinal moments at 7 swing points, including flexion-extension moment, lateral bending moment, and vertical rotation moment at L5/S1 level, between golfers with and without LBP. Statistical significances were considered at the $p<0.0167$ after Bonferroni's correction.

Trunk and Hip Strength

One-tailed dependent t-tests were used to determine significant differences in each trunk muscle strength, including trunk extension, flexion, and bilateral rotation, between golfers with and without LBP. Statistical significances were considered at the $p<0.0125$ after Bonferroni's correction.

Two-tailed dependent t-tests were used to determine significant differences in strength ratios of trunk extension/flexion and right/left rotation at speeds of 60 and 120°/second between golfers with and without LBP (Hypothesis 3.1). Statistical significances were considered at the $p<0.025$ after Bonferroni's correction.

One-tailed dependent t-tests were also used to determine significant differences in each hip muscle strength and side-to-side strength difference of each hip muscle group (Hypothesis 3.2), including hip extensors, flexors, abductors, and adductors, between golfers with and without LBP. Statistical significances were considered at the $p<0.0125$ after Bonferroni's correction.

Trunk and Hip Flexibility

One-tailed dependent t-tests were used to determine significant differences in the ROM of trunk flexion, extension, and right and left rotation between golfers with and without LBP (Hypothesis 4.1). Statistical significances were considered at the $p<0.0125$ after Bonferroni's correction.

One-tailed dependent t-tests were used to determine significant differences in the average ROM of bilateral hip flexion, extension, and hamstrings between golfers with and without LBP (Hypothesis 4.2). Statistical significances were considered at the $p<0.0167$ after Bonferroni's correction.

One-tailed dependent t-tests were used to determine significant differences in each hip ROM between golfers with and without LBP. Statistical significances were considered at the $p<0.0167$ for hip flexion, extension, and hamstrings, $p<0.025$ for hip abduction and adduction, $p<0.025$ for hip internal and external rotation, and $p<0.05$ for FABERE's distance after Bonferroni's correction.

One-tailed dependent t-tests were also used to determine significant differences between lead and non-lead legs in each hip ROM for both groups of the subjects (Hypothesis 4.3). Statistical significances were considered at the $p<0.0167$ for hip flexion, extension, and hamstrings, $p<0.025$ for hip abduction and adduction, $p<0.025$ for hip internal and external rotation, and $p<0.05$ for FABERE's distance after Bonferroni's correction.

Back Proprioception

One-tailed dependent t-tests were used to determine significant differences in active spinal repositioning errors in trunk flexion and extension between golfers with and without LBP (Hypothesis 5.1). Statistical significances were considered at the $p<0.025$

after Bonferroni's correction.

One-tailed dependent t-tests were used to determine significant differences in active spinal repositioning errors in trunk right and left side bending between golfers with and without LBP (Hypothesis 5.2). Statistical significances were considered at the $p<0.025$ after Bonferroni's correction.

One-tailed dependent t-tests were used to determine significant differences in active spinal repositioning errors in trunk right and left rotation between golfers with and without LBP (Hypothesis 5.3). Statistical significances were considered at the $p<0.025$ after Bonferroni's correction.

Postural Stability

Two-way analysis of covariance (ANCOVA) with two within-subject factors was used to determine significant differences in the sway velocity of the COP while standing on one leg with eyes open or eyes closed between golfers with and without LBP (Hypothesis 6). One within-subjects factor was the group (golfers with and without LBP were matched by age, gender, and handicap). Another within-subjects factor was the side of legs (better side: the side with slower COP sway velocity; worse side: the side with faster COP sway velocity) (Luoto et al., 1998). The differences between the two groups were compared separately for the eyes open and eyes closed condition. The dependent variable was the sway velocity of the COP. Subjects' body height was the covariant. Statistical significances were considered at the $p<0.05$.

IV. RESULTS

The purpose of this study was to examine the kinematics and kinetics of the trunk during the golf swing in golfers with and without LBP and to assess their corresponding physical characteristics. The trunk kinematics during the golf swing included maximum X factor normalized by the maximum trunk rotation angle toward the non-lead side in neutral position, maximum crunch factor, maximum incline factor, trunk extension angle at the end of swing, and maximum spinal rotation velocity. The lumbar spinal kinetics during the golf swing included maximum forces along the three anatomical axes and maximum moments about the three anatomical axes at the L5/S1 level. One-tailed dependent t-tests were used to determine significant differences in kinematic and kinetic variables between golfers with and without LBP. The physical characteristics assessed included trunk strength, hip strength, trunk flexibility, hip flexibility, hamstring flexibility, back proprioception, and postural stability. Two-tailed dependent t-tests were used to determine significant differences in trunk strength ratios between the two groups. Two-way analysis of covariance (ANCOVA) with two within-subject factors (two groups of subjects were matched, legs within each subject) was used to determine significant differences in postural stability between golfers with and without LBP. One-tailed dependent t-tests were used for other variables of physical characteristics. Statistical significance was set at $p < 0.05$ for all procedures. To control for inflated Type I error rate due to the number of tests performed, the Bonferroni correction was applied within groupings of related tests. For example if four tests were considered to constitute a grouping, each of the four would be tested at a significance level of 0.05/4. Decisions about which tests were considered to be part of the same grouping were made on the basis of sub-hypotheses within each major hypothesis.

A. Subject Characteristics

Sixteen male golfers with history of golf related mechanical LBP resulting from golf within the past two years were matched by age and handicap to sixteen male golfers with no history of LBP in this study. All golfers were right-handed with USGA handicap lower than 20 and between the ages of 30-60 years old. Demographic data for all subjects are presented in Table 4.1. Golfers in the LBP group experienced back pain localized over the right or central lumbosacral area. Their worst episode of LBP within the past two years had an average score of 45.3 ± 18.2 on the modified Oswestry questionnaire. The disability associated with this modified Oswestry score represents inability to stand for 15 minutes or more, to sit for 30 minutes or more, or to walk for more than 0.25 mile (0.4 km) without increased pain. The LBP golfers were asymptomatic at the time of testing.

Table 4.1 Characteristics of the subjects

	With LBP (n = 16)		Without LBP (n = 16)	
	Mean	SD	Mean	SD
Age (yrs)	48.6	7.4	47.9	8.3
Height (cm)	178.2	5.4	181.4	8.0
Mass (kg)	88.3	18.2	87.5	9.6
Handicap	9.1	4.6	9.5	4.8
Modified Oswestry score	45.3	18.2		

B. Trunk Kinematics During the Golf Swing

The trunk kinematic variables assessed during the golf swing for each group of golfers are presented in Table 4.1 - 4.3 with the variables tested in Hypotheses 1.1 - 1.3 highlighted. There was no significant difference in the maximum X factor normalized by the maximum trunk rotation angle toward non-lead side in neutral position between the two groups (Table 4.2). However, the LBP group had significantly less maximum X factor during the golf swing and significantly less maximum trunk rotation angle toward non-lead side measured actively in neutral standing position than the group without LBP. No statistically significant differences in the maximum lumbar or trunk crunch factor, maximum incline factor, and the angle of trunk extension at the end of swing were found between the two groups (Table 4.3). Furthermore, maximum spinal rotation velocities (lumbar spine rotation and whole trunk rotation) between the two groups were not significantly different (Table 4.4).

Table 4.2 Maximum X factor during the golf swing, maximum trunk right rotation angle, and normalized maximum X factor

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Max. X factor (deg.)	45.75	8.17	50.20	6.15	0.030 *
Max. trunk rotation angle toward non-lead side[1.] (deg.)	44.61	6.12	51.51	6.47	0.003 *
Normalized max. X factor[2.] (%)	104.00	20.62	98.40	13.27	0.159

1. Max. trunk rotation angle was measured actively in neutral standing position
2. Normalized max. X factor = Max. X factor / Max. trunk rotation angle toward non-lead side x 100

* $p < 0.05$ (Bonferroni's correction)

Table 4.3 Maximum crunch factor, Maximum incline factor, and maximum trunk extension angle during the end of swing

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Max. lumbar crunch factor	2000.34	763.22	2354.57	752.88	0.063
Max. trunk crunch factor	7291.31	2717.99	8686.54	2245.68	0.058
Max. incline factor	533.22	159.31	559.18	158.69	0.306
Max. trunk extension angle (deg.) during the end of swing	2.36	8.10	4.47	7.05	0.175

Table 4.4 Maximum spinal rotation velocity

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Max. spinal rotation velocity (deg./sec.)					
Lumbar	217.64	43.12	217.23	53.99	0.491
Whole trunk	404.14	69.83	437.04	102.60	0.107

C. Lumbar Spinal Kinetics During the Golf Swing

The maximum spinal forces and moments at the L5/S1 level during the golf swing for each group of golfers are presented in Table 4.5 with the variables tested in Hypotheses 2.1 – 2.2 highlighted. No significant differences were found in maximum anterior shear force, posterior shear force, lateral shear force, and compression force at the L5/S1 level between the two groups. No significant differences were found in the maximum moments about the three anatomical axes at the L5/S1 level between the two groups. Furthermore, the spinal forces and moments at seven swing points of the golf swing were also compared between the golfers with and without LBP (Figure 4.1 - 4.6). No significant differences were found in the spinal forces and moments at any analyzed swing point between the two groups.

Table 4.5 Maximum spinal forces and moments (at L5/S1) during the golf swing

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Spinal force in each direction of anatomical axis					
Max. shear force to right side (%BW)	15.57	4.50	15.13	4.35	0.347
Max. shear force to left side (%BW)	-5.66	4.27	-5.50	4.25	0.451
Max. anterior shear force (%BW)	19.44	18.63	15.06	10.42	0.252
Max. posterior shear force (%BW)	-13.43	15.28	-15.56	10.11	0.340
Max. compression force (%BW)	-88.93	12.62	-91.98	13.14	0.287
Spinal moment in each anatomical plane					
Max. flexion moment (%BW*HT)	-7.94	4.35	-7.53	4.34	0.392
Max. extension moment (%BW*HT)	0.93	2.22	0.12	1.19	0.091
Max. left bending moment (%BW*HT)	-5.16	3.33	-4.41	2.45	0.215
Max. right bending moment (%BW*HT)	4.37	2.39	4.94	2.90	0.245
Max. back rotation moment (%BW*HT)	-3.84	2.17	-4.33	2.37	0.273
Max. forward rotation moment (%BW*HT)	1.73	1.05	1.51	0.88	0.246

%BW = % body weight
%BW*BH = % body weight * height

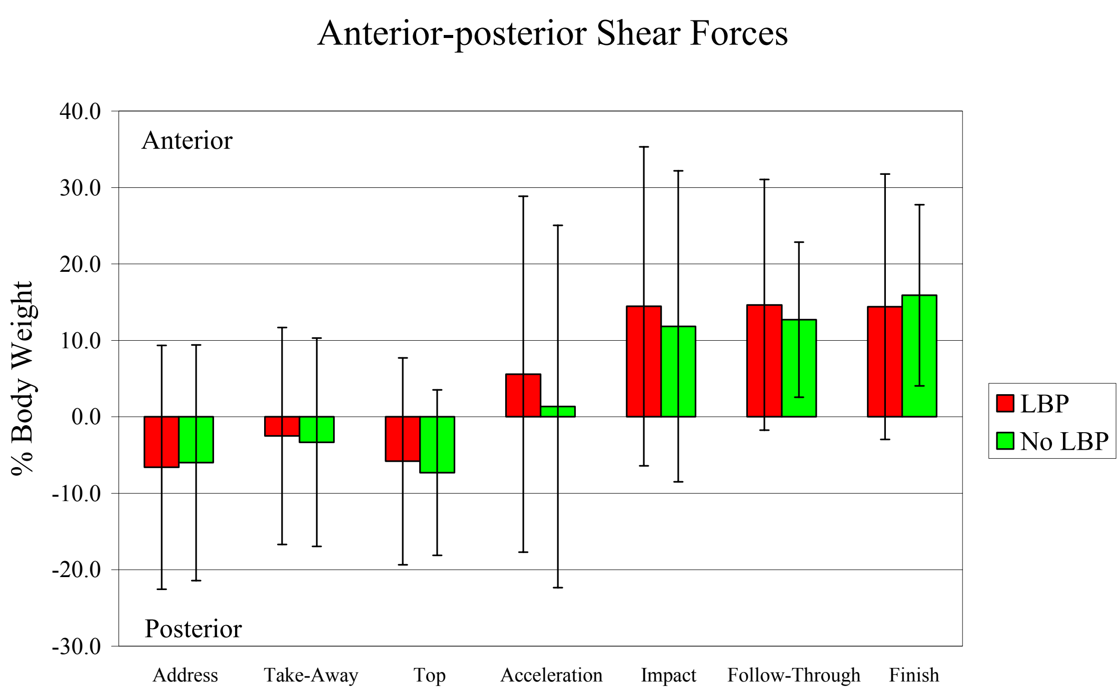


Figure 4.1 Anterior-posterior shear forces (at L5/S1) at 7 swing points

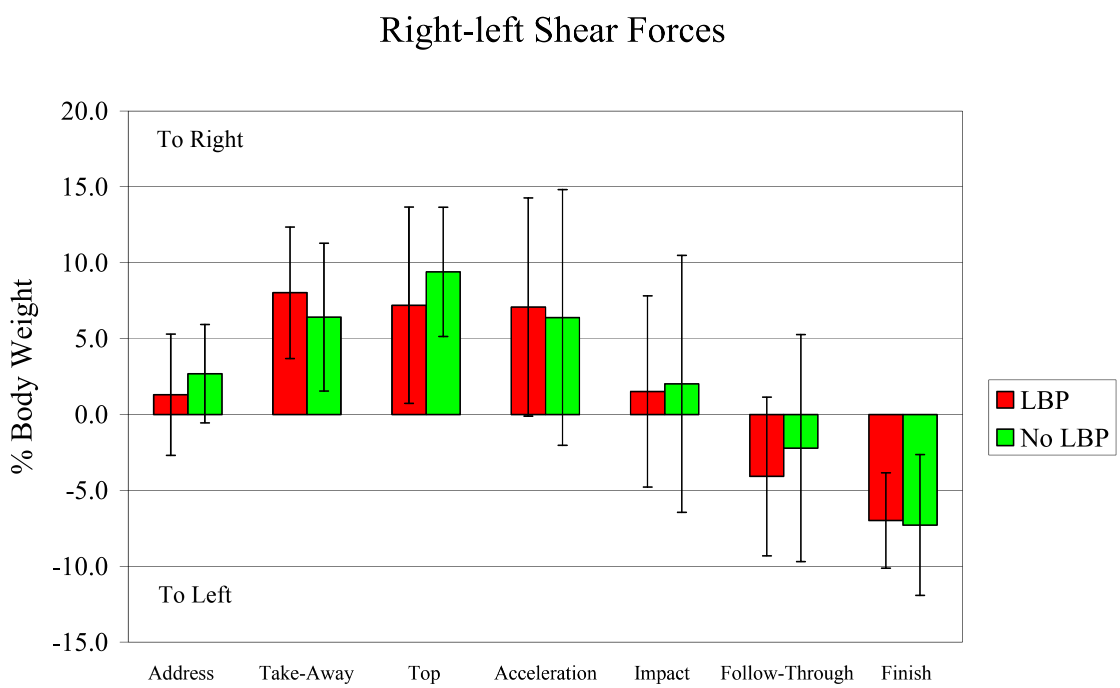


Figure 4.2 Right-left shear forces (at L5/S1) at 7 swing points

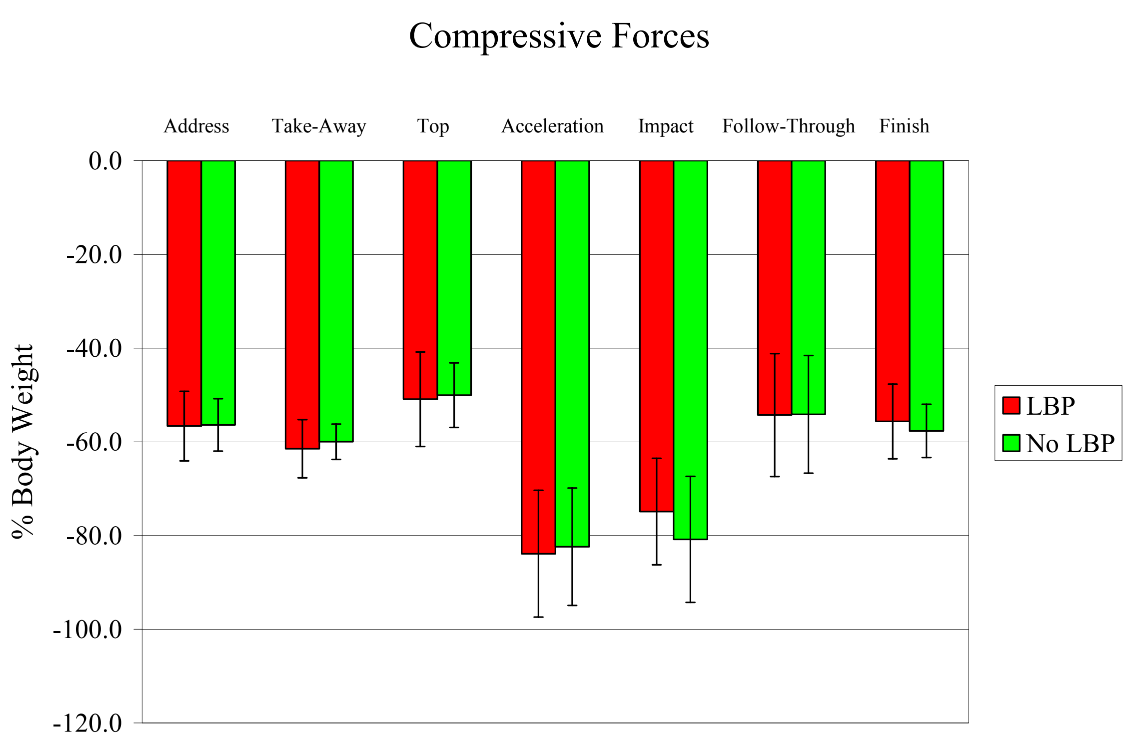


Figure 4.3 Compressive forces (at L5/S1) at 7 swing points

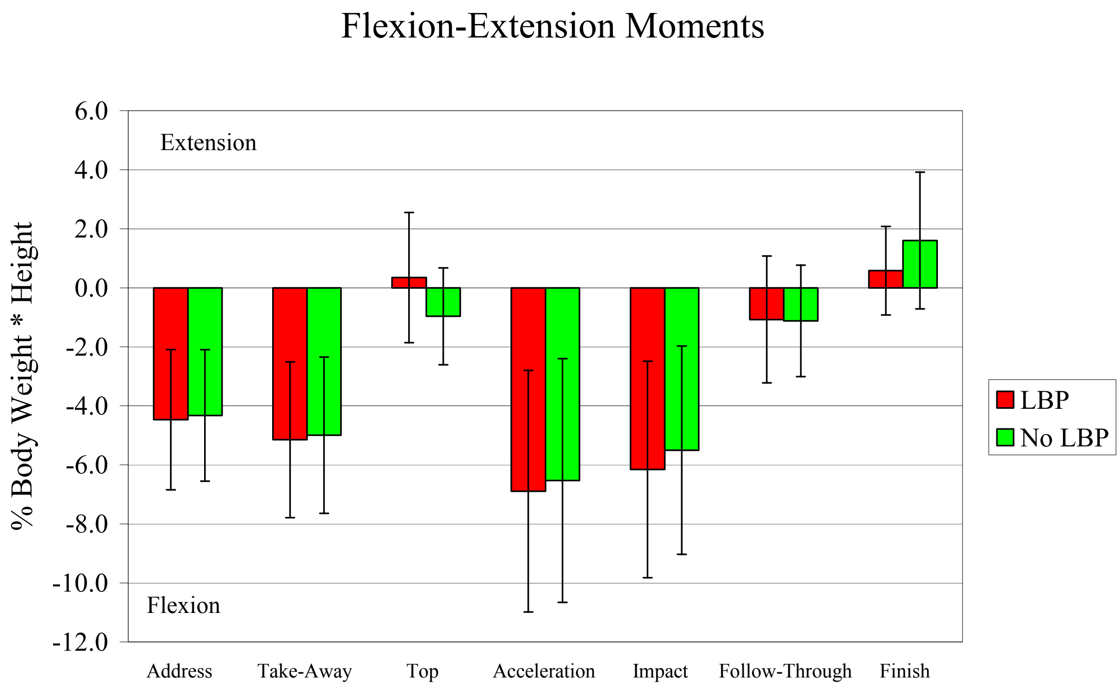


Figure 4.4 Flexion-extension moments (at L5/S1) at 7 swing points

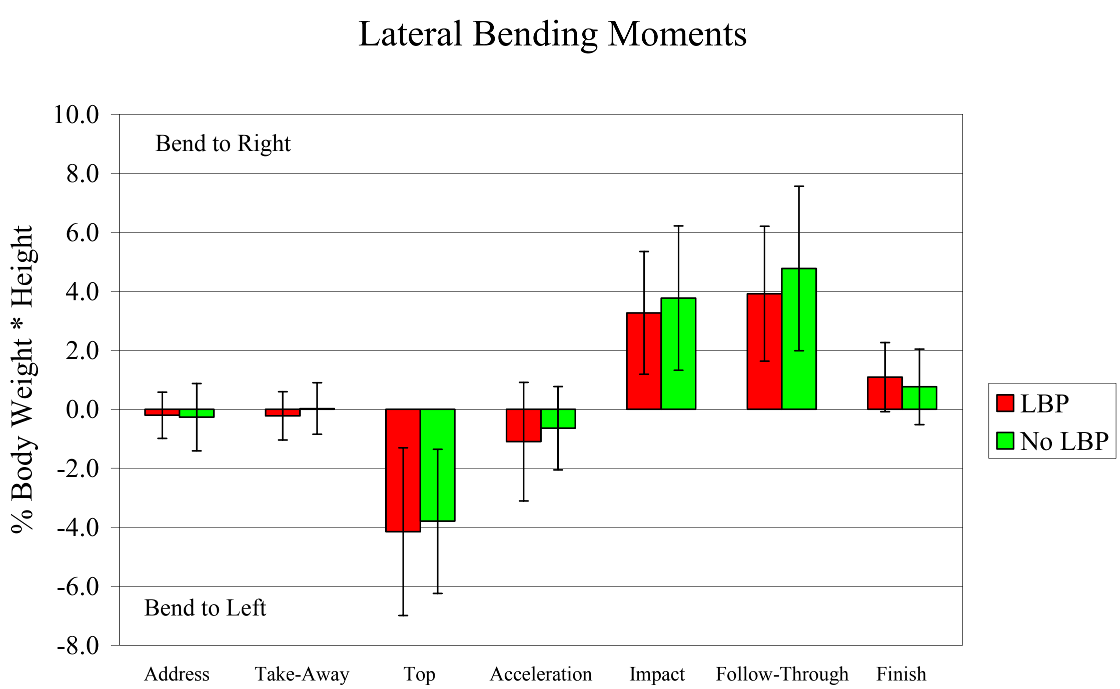


Figure 4.5 Lateral bending moments (at L5/S1) at 7 swing points

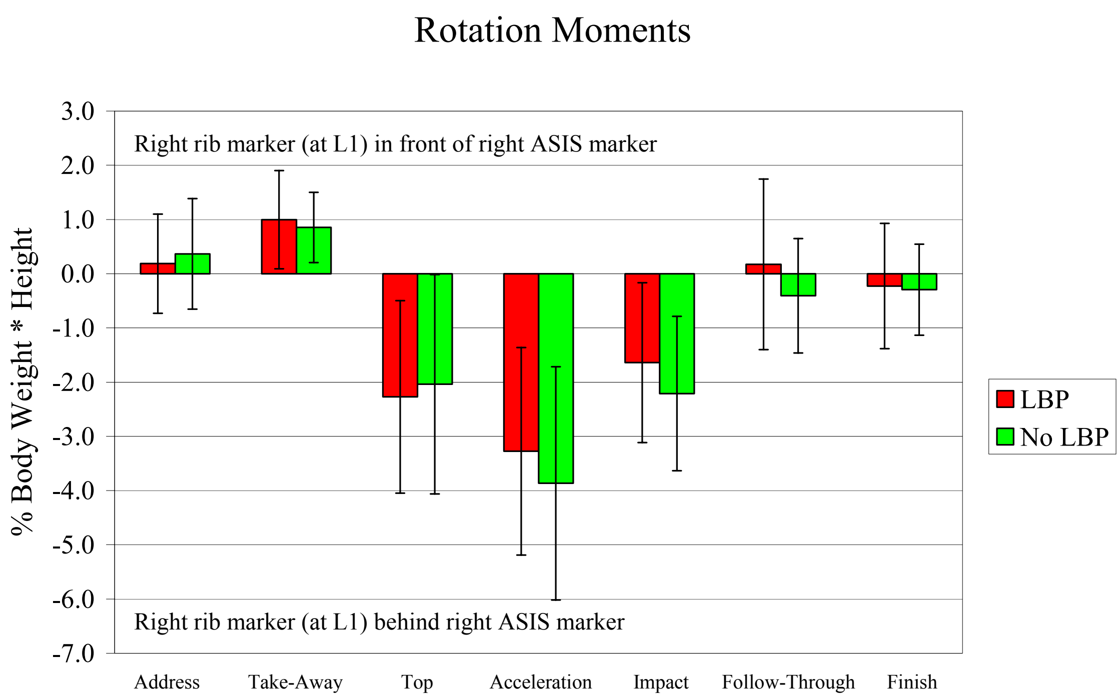


Figure 4.6 Rotation moments (at L5/S1) at 7 swing points

D. Trunk and Hip Strength

Isokinetic trunk strength and strength ratios of trunk extension/flexion and right/left rotation at speeds of 60 and 120 degrees/sec for each group of golfers are presented in Table 4.6 with the variables tested in Hypothesis 3.1 highlighted. Statistical significance was set at $p < 0.025$ for strength ratios and $p < 0.0125$ for individual strength measurements after Bonferroni correction. The LBP group demonstrated significantly less trunk extension/flexion strength ratios than the group without LBP at the speed of 60 degrees/sec (1.47 ± 0.26 vs. 1.75 ± 0.32, $t = -3.06$, $p = 0.008$). The LBP group demonstrated significantly less trunk extension strength than the group without LBP at 60 degrees/sec ($285.61 \pm 56.11\%$ vs. $361.92 \pm 86.92\%$ body weight, $t = -5.53$, $p < 0.001$). No significant difference was found in trunk flexion strength at 60 degrees/sec between the two groups. In addition, the LBP group did not demonstrate a significant difference in the strength ratio of trunk extension/flexion at the speed of 120 degrees/sec compared to the group without LBP. There were also no significant differences in trunk extension strength at 120 degrees/sec and trunk flexion strength at 60 or 120 degrees/sec between the two groups.

No significant differences were found in the strength ratios of trunk rotation at both speeds between the two groups. However, the LBP group demonstrated significantly less left trunk rotation strength (122.28 ± 29.77 vs. 146.06 ± 26.40 % body weight, t = -3.61, p = 0.001) than the group without LBP at 60 degrees/sec. There was no significant difference in right trunk rotation strength at 60 degrees/sec between the two groups. There were also no significant differences in bilateral trunk rotation strength at the speed of 120 degrees/sec between the two groups.

Table 4.6 Isokinetic strength of each trunk muscle group and strength ratios at different speeds

		With LBP		Without LBP		Paired T-Test	
						One-tailed	Two-tailed
		Mean	SD	Mean	SD	P value	P value
60 deg/sec	Extension (%BW)	285.61	56.11	361.92	86.92	0.000 **	
	Flexion (%BW)	196.10	33.81	208.45	41.00	0.138	
	E/F ratio	1.47	0.26	1.75	0.32		0.008 *
	Right rotation (%BW)	121.93	36.60	141.72	26.77	0.021	
	Left rotation (%BW)	122.28	29.77	146.06	26.40	0.001 **	
	R/L rotation ratio	0.99	0.14	0.98	0.13		0.840
120 deg/sec	Extension (%BW)	316.68	73.46	351.72	88.53	0.039	
	Flexion (%BW)	169.59	38.12	181.92	34.85	0.143	
	E/F ratio	1.90	0.38	1.96	0.46		0.636
	Right rotation (%BW)	127.73	40.13	140.28	26.37	0.132	
	Left rotation (%BW)	124.89	31.69	139.53	21.45	0.045	
	R/L rotation ratio	1.01	0.15	1.00	0.10		0.804

Note:
E/F ratio = Trunk extension strength / flexion strength
R/L rotation ratio = Trunk right rotation strength / left rotation strength
%BW = % body weight
* p<.025 (Bonferroni correction)
** p<.0125 (Bonferroni correction)

Isometric hip strength and side-to-side strength differences of each hip muscle group for each group of golfers are presented in Table 4.7 and 4.8 with the variables tested in Hypothesis 3.2 highlighted. Statistical significance was set at $p < 0.0125$ for individual strength measurements and side-to-side strength differences after Bonferroni correction. No significant differences were found in side-to-side strength of the hip abductors, adductors, flexors, and extensors between the two groups. However, the LBP group demonstrated significantly less left hip abduction strength (132.62 ± 37.56% vs. 157.14 ± 29.10% body weight, $t = -2.64$, $p = 0.009$), less right hip flexion strength (63.60 ± 24.13% vs. 88.32 ± 37.00% body weight, $t = -2.53$, $p = 0.012$), and less left hip flexion strength (58.21 ± 28.05% vs. 80.08 ± 28.50% body weight, $t = -2.83$, $p = 0.006$) than the group without LBP. No significant differences were found in bilateral hip adduction and bilateral extension strength between the two groups.

Table 4.7 Isometric strength of each hip muscle group

		With LBP		Without LBP		Paired T-Test One-tailed
		Mean	SD	Mean	SD	P value
Abduction	Right leg (%BW)	135.93	42.54	150.19	23.42	0.092
	Left leg (%BW)	132.62	37.56	157.14	29.10	0.009 *
Adduction	Right leg (%BW)	124.03	34.34	152.42	45.48	0.024
	Left leg (%BW)	124.22	35.97	162.21	42.21	0.020
Extension	Right leg (%BW)	262.09	77.10	316.45	49.06	0.031
	Left leg (%BW)	269.81	70.99	327.30	60.43	0.032
Flexion	Right leg (%BW)	63.60	24.13	88.32	37.00	0.012 *
	Left leg (%BW)	58.21	28.05	80.08	28.50	0.006 *

%BW = % body weight
* p<.0125 (Bonferroni correction)

Table 4.8 Side-to-side strength difference of each hip muscle group

	With LBP		Without LBP		Paired T-Test One-tailed
Side to side difference (%)	Mean	SD	Mean	SD	P value
Abduction	13.38	7.98	10.81	8.10	0.155
Adduction	21.48	11.48	15.49	10.60	0.074
Extension	11.23	9.55	12.56	7.01	0.323
Flexion	25.63	16.50	20.95	12.90	0.184

E. Trunk and Hip Flexibility

Active range of motion (ROM) of trunk movements for each group of golfers are presented in Table 4.9 with the variables tested in Hypothesis 4.1 highlighted. Statistical significance was set at $p < 0.0125$ for each trunk ROM after Bonferroni correction. The LBP group demonstrated significantly less ROM than the group without LBP in trunk right rotation (43.24 ± 4.67° vs. 46.55 ± 6.58°, $t = -2.72$, $p = 0.008$). There were no significant differences in the ROM of trunk flexion, extension, and left rotation between the two groups.

Table 4.9 Active range of motion of trunk movements

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Flexion (deg.)	56.20	12.33	54.56	11.14	0.349
Extension (deg.)	25.94	6.80	28.35	8.14	0.199
Right rotation (deg.)	43.24	4.67	46.55	6.58	0.008 *
Left rotation (deg.)	40.64	7.64	43.40	8.03	0.161

* p<.0125 (Bonferroni correction)

Hip ROM and knee flexion angles during active knee extension test for each group of golfers are presented in Table 4.10 and Table 4.11 with the variables tested in Hypotheses 4.2 and 4.3 highlighted. Table 4.10 includes comparisons of hip ROM between golfers with and without LBP. Table 4.11 includes comparisons of bilateral hip ROM within each group of golfers. With Bonferroni correction, statistical significance was set at $p < 0.0167$ for the average ROM of bilateral hip flexion, the average ROM of bilateral hip extension, and the average of bilateral knee flexion angle during active knee extension test. Statistical significance was also set at $p < 0.0167$ for each leg's hip flexion ROM, hip extension ROM, and knee flexion angle during active knee extension test. In addition, statistical significance was set at $p < 0.025$ for the ROM of hip abduction, adduction, internal rotation, and external rotation for each leg. Statistical significance was set at $p < 0.05$ for FABERE's distance.

Table 4.10 shows that the mean bilateral knee flexion angles during active knee extension test were greater in the LBP group compared to the group without LBP (24.13 ± 8.78° vs. 17.44 ± 6.63°, $t = 2.68$, $p = 0.008$). The LBP group demonstrated greater right knee flexion angle (22.09 ± 8.03° vs. 16.29 ± 6.25°, $t = 2.45$, $p = 0.013$) and left knee flexion angle (26.18 ± 10.49° vs. 18.59 ± 7.47°, $t = 2.60$, $p = 0.010$) during the active knee extension test than the group without LBP. There were no significant differences in the average ROM of bilateral hip flexion and extension between the two groups. No significant differences were found in the ROM of hip flexion, extension, abduction, adduction, internal rotation, external rotation, and FABERE's distance for each leg between the two groups as well.

Table 4.10 Comparisons of Hip ROM between golfers with and without LBP

		With LBP		Without LBP		Paired T-Test One-tailed
		Mean	SD	Mean	SD	P value
Flexion (deg.)	Right leg	133.63	9.05	139.75	6.19	0.036
	Left leg	133.75	9.17	136.63	5.67	0.149
	Mean	133.69	8.89	138.19	5.55	0.065
Extension (deg.)	Right leg	16.56	5.11	18.25	5.52	0.217
	Left leg	20.13	3.26	19.81	4.61	0.394
	Mean	18.34	3.58	19.03	4.74	0.324
Knee flexion angle (deg.) during active knee extension test	Right leg	22.09	8.03	16.29	6.25	0.013 *
	Left leg	26.18	10.49	18.59	7.47	0.010 *
	Mean	24.13	8.78	17.44	6.63	0.008 *
Abduction (deg.)	Right leg	28.31	5.94	30.63	5.76	0.066
	Left leg	29.88	6.40	33.25	6.81	0.081
Adduction (deg.)	Right leg	15.13	4.29	17.13	3.01	0.058
	Left leg	14.63	3.42	16.50	4.56	0.107
Internal rotation (deg.)	Right leg	37.56	8.20	40.81	7.30	0.127
	Left leg	38.81	8.00	39.00	8.21	0.477
External Rotation (deg.)	Right leg	32.63	8.07	37.69	8.75	0.070
	Left leg	35.00	8.06	39.06	7.93	0.105
FABERE's distance (cm)	Right leg	18.04	4.14	17.27	4.28	0.323
	Left leg	19.89	5.18	18.47	4.51	0.191

* $p<.0167$ (Bonferroni correction)

Table 4.11 shows that the LBP group demonstrated less right hip external rotation ROM than left hip (32.63 ± 8.07° vs. 35.00 ± 8.06°, t = 2.29, p = 0.018). The LBP group also demonstrated less right hip extension ROM than left hip (16.56 ± 5.11° vs. 20.13 ± 3.26°, t = -3.02, p = 0.004). Additionally, the LBP group demonstrated less right knee flexion angle than left knee during active knee extension test (22.09 ± 8.03° vs. 26.18 ± 10.49°, t = 2.56, p = 0.011). On the other hand, the group without LBP demonstrated less left hip flexion ROM than right hip (136.63 ± 5.67° vs. 139.75 ± 6.19°, t = 2.96, p = 0.005). The group without LBP also demonstrated less right knee flexion angle than left knee during active knee extension test (16.29 ± 6.25° vs. 18.59 ± 7.47°, t = 2.46, p = 0.013). Other hip ROMs were similar between legs within each group of golfers.

Table 4.11 Comparisons of bilateral hip ROM within each group of golfers

		Left leg (Lead side) Mean	Left leg (Lead side) SD	Right leg (Non-lead side) Mean	Right leg (Non-lead side) SD	Paired T-Test One-tailed P value
With LBP	Abduction (deg.)	29.88	6.40	28.31	5.94	0.101
	Adduction (deg.)	14.63	3.42	15.13	4.29	0.300
	Internal rotation (deg.)	38.81	8.00	37.56	8.20	0.161
	External rotation (deg.)	35.00	8.06	32.63	8.07	0.018 *
	FABERE's distance (cm)	19.89	5.18	18.04	4.14	0.062
Without LBP	Abduction (deg.)	33.25	6.81	30.63	5.76	0.029
	Adduction (deg.)	16.50	4.56	17.13	3.01	0.297
	Internal rotation (deg.)	39.00	8.21	40.81	7.30	0.109
	External rotation (deg.)	39.06	7.93	37.69	8.75	0.176
	FABERE's distance (cm)	18.47	4.51	17.27	4.28	0.088
With LBP	Flexion (deg.)	133.75	9.17	133.63	9.05	0.451
	Extension (deg.)	20.13	3.26	16.56	5.11	0.004 **
	Knee flexion angle (deg.) during active knee extension test	26.18	10.49	22.09	8.03	0.011 **
Without LBP	Flexion (deg.)	136.63	5.67	139.75	6.19	0.005 **
	Extension (deg.)	19.81	4.61	18.25	5.52	0.055
	Knee flexion angle (deg.) during active knee extension test	18.59	7.47	16.29	6.25	0.013 **

* p<.025 (Bonferroni correction)
** p<.0167 (Bonferroni correction)
Note: All golfers were right handed.

F. Back Proprioception

Active spinal repositioning errors in three anatomical planes for each group of golfers are presented in Table 4.12 with the variables tested in Hypotheses 5.1 - 5.3 highlighted. Statistical significance was set at $p < 0.025$ for each direction of trunk movements after Bonferroni correction. The LBP group demonstrated significantly greater spinal repositioning errors than the group without LBP in trunk flexion ($3.24 \pm 1.46°$ vs. $2.13 \pm 0.86°$, $t = 3.11$, $p<0.025$). There were no statistically significant differences in spinal repositioning error for trunk extension, right and left rotation and side bending between the two groups.

Table 4.12 Active spinal repositioning errors in three anatomical planes

	With LBP		Without LBP		Paired T-Test One-tailed
	Mean	SD	Mean	SD	P value
Flexion (deg.)	3.24	1.46	2.13	0.86	0.004 *
Extension (deg.)	2.02	1.20	1.94	0.91	0.407
Right rotation (deg.)	2.91	1.27	2.22	0.68	0.034
Left rotation (deg.)	2.76	1.57	2.51	0.83	0.258
Right bending (deg.)	2.17	1.03	1.57	0.52	0.035
Left bending (deg.)	2.12	1.09	1.73	0.67	0.129

* $p<.025$ (Bonferroni correction)

G. Postural Stability

Descriptive statistics and the summaries of two-way analysis of covariance (ANCOVA) for differences in postural stability assessment with eyes open or eyes closed between golfers with and without LBP are presented in Tables 4.13 and 4.14. No significant differences were found in the sway velocity of the COP while standing on one leg with either eyes open or eyes closed between golfers with and without LBP.

Table 4.13 Sway velocity of the COP while standing on one leg with eyes open or eyes closed

		With LBP		Without LBP	
	Legs	Mean	SD	Mean	SD
Eyes open	Better side	4.05	1.38	4.73	1.44
	Worse side	5.36	2.34	5.87	2.21
Eyes closed	Better side	9.48	3.14	10.09	3.45
	Worse side	11.37	3.70	12.10	3.70

Note:
Unit of sway velocity of the center of pressure (COP) is cm/sec
Better side is the side with slower COP sway velocity
Worse side is the side with faster COP sway velocity

Table 4.14 Summary of two-way ANCOVA for the differences in sway velocity of the COP while standing on one leg with eyes open and eyes closed between golfers with and without LBP

	Legs	With LBP Adjusted Mean	Without LBP Adjusted Mean
Eyes open	Better side	4.10	4.68
	Worse side	5.41	5.82
Eyes closed	Better side	9.71	9.86
	Worse side	11.60	11.97

Note:
Unit of sway velocity of the center of pressure (COP) is cm/sec
Better side is the side with slower COP sway velocity
Worse side is the side with faster COP sway velocity

With eyes open

Source	SS	df	MS	F	P
Adjusted Velocities	23.89	1	23.89	17.74	0.001
Adjusted Groups	0.30	1	0.30	0.07	0.798
Adjusted Velocities x Groups	0.10	1	0.10	0.12	0.732

With eyes closed

Source	SS	df	MS	F	P
Adjusted Velocities	60.65	1	60.65	87.00	0.000
Adjusted Groups	0.54	1	0.54	0.00	0.966
Adjusted Velocities x Groups	0.06	1	0.06	0.09	0.769

Note:
Velocities - one leg with slower and one leg with faster sway velocity of the COP
Groups - golfers with and without LBP

V. DISCUSSION

Sub-optimal physical fitness and improper swing mechanics may produce abnormal forces to the lumbar spine and may contribute to the development of mechanical LBP (Geisler, 2001; Hosea & Gatt, 1996; McCarroll & Gioe, 1982; Stover et al., 1976). Physical characteristics that are associated with LBP include trunk and hip muscle weakness and strength imbalance (Andersson et al., 1988; Beimborn & Morrissey, 1988; Davis & Marras, 2000; J. H. Lee et al., 1999; McNeill et al., 1980), trunk and hip flexibility deficits (Ashmen et al., 1996; Kendall et al., 1993; Mellin, 1987, 1988; Trainor & Trainor, 2004; Vad et al., 2004), deficient back proprioception (Brumagne et al., 2000; Gill & Callaghan, 1998; Laskowski et al., 2000; Newcomer, Laskowski, Yu, Larson et al., 2000), and poor postural control (Luoto et al., 1998; Mientjes & Frank, 1999). Excessive trunk rotation during the backswing (Lindsay & Horton, 2002), crunch factor (Morgan et al., 1997), spinal rotation velocity (Hosea et al., 1990), trunk hyperextension at the end of swing (Fischer & Watkins, 1996; Geisler, 2001), and incline factor (evaluated in this study) are potential swing mechanics that may contribute to low back injuries. However, differences in physical characteristics, trunk motion, and the resulting spinal loads during the golf swing between golfers with and without mechanical LBP remain unclear. The purpose of this study was to examine the physical characteristics (including trunk and hip strength, trunk and hip flexibility, back proprioception, and postural stability) and the trunk kinematics and kinetics during the golf swing in golfers with and without LBP.

The results of this study revealed that golfers with LBP demonstrated less trunk rotation strength, less trunk extension strength, and smaller strength ratio in trunk extension/flexion. The LBP group also demonstrated less hip muscle strength. Additionally, the LBP group had less trunk rotation ROM toward non-lead side and hamstring flexibility. The LBP group also demonstrated a significant proprioception deficit in trunk flexion. However, no statistically significant differences were found in the trunk kinematics and the spinal loads investigated in this study between the two groups. Results according to the specific aims of this study and the relevant findings will be discussed.

A. Pain Status of the Low Back Pain Group

Golfers in the LBP group reported an average modified Oswestry score of 45.3 ± 18.2 based on their worst episodes during the two years prior to testing. Generally, individuals with LBP who have a modified Oswestry score between 40 and 60 are usually unable to stand for 15 minutes or more, to sit for 30 minutes or more, or to walk for more than 0.25 mile (0.4 km) without increased pain (Delitto, Erhard, & Bowling, 1995). Individuals with LBP who have a modified Oswestry score between 20 and 40 are usually able to sit, stand, or walk, but pain prevents them from performing activities of daily living, such as vacuuming, lifting, and mowing the grass (Delitto et al., 1995). The results of this study demonstrated that golfers in the LBP group experienced golf related mechanical LBP that resulted in their inability to perform basic mechanical functions (standing, walking, or sitting) or activities of daily living. It was considered that if an individual is unable to perform these fundamental activities, the person can not be expected to perform a more complex and stressful activity (Delitto et al., 1995), such as golf. This reveals that LBP had significant impact on the ability to play golf in golfers with LBP in the current study.

As reported in the literature, LBP is the most common injury among golfers (Batt, 1992; Duda, 1987; Gosheger et al., 2003; McCarroll et al., 1990). More than 20% professional golfers and 30% amateur golfers have experienced low back injury (Batt, 1992; Gosheger et al., 2003; McCarroll et al., 1990). According to a survey, low back injuries may result in disability, which could inhibit golf participation on average for ten weeks (Gosheger et al., 2003). It was reported that 16% professional golfers and 31% amateur golfers who suffered injuries were treated with rest alone (McCarroll & Gioe, 1982; McCarroll et al., 1990). It is essential for golfers and clinicians to understand the deficits that golfers with LBP may have and provide appropriate training exercises for injury prevention and rehabilitation.

B. Physical Characteristics

1. Trunk and Hip Strength

Strength imbalances between the trunk muscles have been reported to be associated with LBP as the trunk muscles provide the mechanical stability of the spine during activities (Andersson et al., 1988; J. H. Lee et al., 1999). The most common cited strength ratio of trunk extensor to flexor for healthy individuals is 1.3 : 1 (Beimborn & Morrissey, 1988). Individuals with LBP, however, were reported to have trunk strength ratios from 0.79 to 1.23 for extension/flexion (Beimborn & Morrissey, 1988). Golfers with LBP in the current study demonstrated smaller isokinetic trunk strength ratios than golfers without LBP at the speed of 60 degrees/sec (1.47 ± 0.26 vs. 1.75 ± 0.32). Further analysis found that the difference in the trunk extension/flexion ratios between the two groups was because golfers with LBP had significantly less trunk extension strength. During the golf swing, a flexed trunk angle must be maintained to make a proper turn back and return to the ball (Adlington, 1996). This positioning requires strong back extensor muscles to support the upper body, especially during the golf swing as rapid and powerful movements can generate considerable spinal loads. Weakness of the back extensor muscles may not generate sufficient strength to counteract the flexion moment produced by the abdominal muscles. Since the sample of swings tested in the study was low and in the laboratory environment, the results can not be extrapolated to an entire round of golf or an exhaustive practice routine on a range where hundreds of swings often occur. It is plausible that after a number of repetitive golf swings, these flexion moments may result in excessive loading in the absence of sufficient trunk extensor muscle strength and overuse injury to the back may occur.

It has been reported that trunk strength ratios are different between non-athletes and athletes and also different among athletes who play different sports (Andersson et al., 1988). Strength ratios of trunk extension/flexion for both groups of golfers in this study were greater than the previously reported ratios (1.3 : 1) for healthy individuals. This difference may be due to two reasons. Golfers may need stronger back muscle strength than non-golfers for performing efficient movements and maintaining trunk stability during the golf swing. Stronger back muscles in golfers may be related to long-term systematic training specific to the golf motion. Another reason may be due to the testing position. The trunk strength and ratios reported in the literature were tested in standing or sitting position (Davies & Gould, 1982; Langrana & Lee, 1984; Langrana, Lee, Alexander et al., 1984; Smith, Mayer, Gatchel et al., 1985; Thompson, Gould, Davies et al., 1985). Trunk extension and flexion strength were tested in semi-standing position in this study in order to simulate the semi-standing position during the golf swing. This position may allow subjects to push with their feet to generate greater trunk extension strength. Further research may be needed to determine if the testing position alters trunk extension torque results. However, it was reported that golfers in general have stronger back extensor muscles when compared to non-athletes of similar age (Weishaupt, Obermuller, & Hofmann, 2000).

Research has suggested that rapid trunk rotation is a potential risk factor associated with LBP (Manning et al., 1984; Marras et al., 1993). Adequate trunk rotation strength may be able to overcome the passive resistance produced by the movements with rotation to prevent low back injury. The results of this study showed that the trunk strength ratios of right to left rotation were similar for both groups. However, the magnitude of trunk rotation strength when normalized to body weight in golfers with LBP was generally weaker than the golfers without LBP.

The hip muscles play a significant role in transferring forces from the lower extremities to the spine during sports activities (Lyons et al., 1983; Nadler et al., 2000). They also assist in maintaining stability of the pelvis and trunk (D. Lee, 1999; Lyons et al., 1983). Research has shown that strength deficits between hips may be a potential risk factor for LBP by decreasing normal lumbo-pelvic-hip stability and contribute to low back injury (Kisner & Colby, 1990; Nadler et al., 2002; Nadler et al., 2000; Nadler et al., 2001). Although the results of this study did not show significant bilateral strength differences of each hip muscle group between golfers with and without LBP, a general weakness of the hip muscles was observed in the LBP group. Weak hip muscles may not be able to maintain stability of pelvis and trunk well during the rapid golf swing in the LBP group.

2. Trunk and Hip Flexibility

Appropriate flexibility of the trunk and hip is critical to prevent low back injuries. Vad et al. reported that golfers with LBP had decreased lumbar extension, decreased lead hip internal rotation, and increased FABERE's distance of the lead hip (Vad et al., 2004). The authors concluded that the limitation in lumbar extension may be a protective mechanism to decrease spinal loads and prevent further symptomatic exacerbation. The lead hip range of motion deficits may result from capsular contractures and subsequent rotation deficits over time as the lead hip acts as the primary pivot point and experiences a significant amount of force. However, similar findings were not observed in this study. Participants in the study of Vad et al. were all professional golfers with existing LBP (Vad et al., 2004). Participants in this study were all amateur golfers without current low back symptoms. It is not known whether the time or frequency of participation in golf activity between professional and amateur golfers resulted in the conflicting findings between the two studies.

In the current study, the LBP group demonstrated significantly less ROM in right trunk rotation in a neutral standing position. Lindsay and Horton reported similar findings in a group of LBP golfers (Lindsay & Horton, 2002). With limited ROM of right trunk rotation, golfers with LBP may not be able to generate a desirably large X factor. In situations where these golfers attempt to generate greater than normal power in their swing to increase the transfer of energy from the club to the ball they may rotate the upper body beyond their physical limitation, placing excessive stress on the back and result in soft tissue injury. The physical limitation of trunk rotation was defined as the maximum ROM of trunk rotation that an individual can reach when rotating from a neutral standing position and at a relatively slow and steady movement speed (Lindsay & Horton, 2002).

In addition, the LBP group in the current study was less flexible in their hamstrings. Studies have also shown tight hamstrings in individuals with LBP who are not golfers (Hultman, Saraste, & Ohlsen, 1992; Mellin, 1988; Nourbakhsh & Arab, 2002). Some authors have suggested that due to the origin of the hamstrings on the ischial tuberosity, decreased flexibility results in decreased lumbar lordorsis by limiting the ability to anteriorly tilt the pelvis and may alter the nucleus pulposus within the disc that can further result in LBP (Kendall et al., 1993; Neumann, 2002a). However, several studies demonstrated that there is no relationship between hamstring flexibility and pelvic tilt or the size of lumbar lordosis (Gajdosik, Hatcher, & Whitsell, 1992; Nourbakhsh & Arab, 2002). Studies also indicated that LBP is not associated with the pelvic tilt or the size of lumbar lordosis (Beninato, Hudson, & Price, 1993; Hansson, Bigos, Beecher et al., 1985; Nourbakhsh & Arab, 2002). It was suggested that hamstring tightness in individuals with LBP is a compensatory mechanism secondary to pelvic instability (van Wingerden, Vleeming, Kleinrensink et al., 1997). In the current study, the LBP group demonstrated tighter hamstrings and weaker hip muscles than the group without LBP. Further analysis to probe this finding which was not originally hypothesized found that the LBP group showed no differences in pelvic tilt during the golf swing when compared to the group without LBP (Figure 5.1). Whether tight hamstring is a result of pelvic instability or a potential risk factor that contributed to the back injuries needs further investigation.

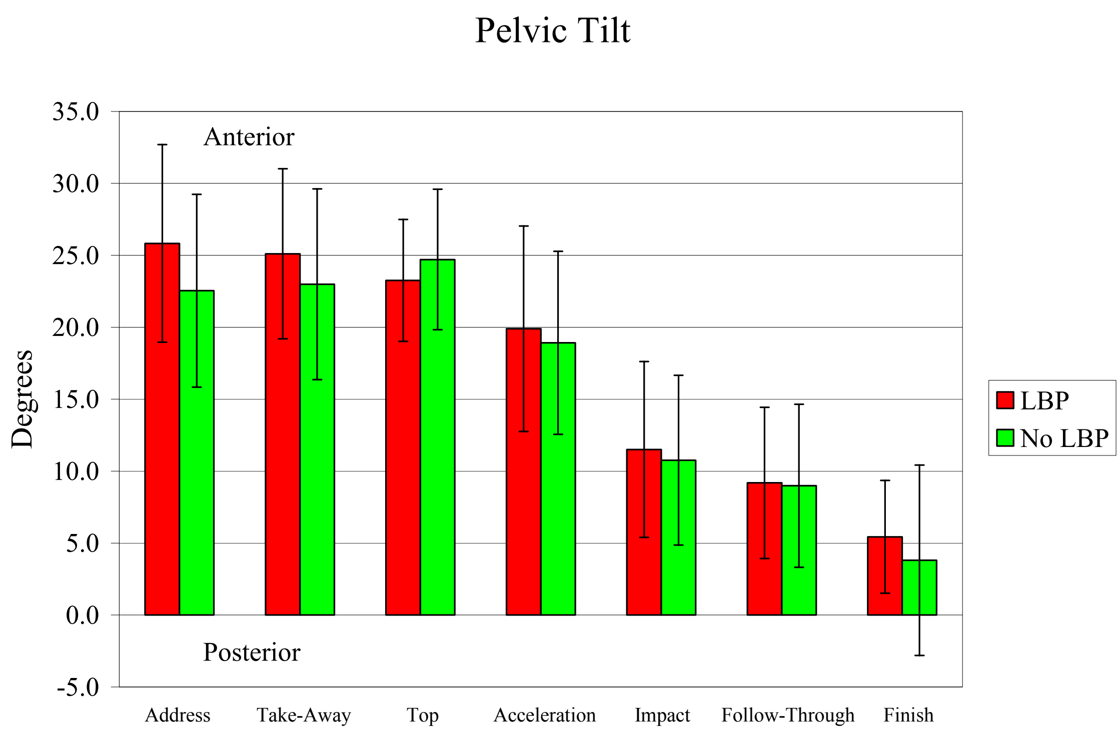


Figure 5.1 Pelvic tilt angle at 7 swing points

3. <u>Back Proprioception</u>

Back injuries may result in proprioceptive deficits due to mechanoreceptor dysfuction (Laskowski et al., 2000; Yamashita et al., 1990). Dysfunction in the neural control system of the lumbar spine may place other spinal structures at risk for injury and also alter the spinal stabilization system (Panjabi, 1992). Research has suggested that individuals with LBP present proprioceptive deficits in trunk flexion (Brumagne et al., 2000; Gill & Callaghan, 1998; Newcomer, Laskowski, Yu, Johnson et al., 2000). Those individuals were tested while symptomatic for their chronic LBP. In the current study, the LBP group although asymptomatic still demonstrated significant proprioceptive deficits in trunk flexion. Such proprioception deficits may influence motor programming for neuromuscular control and muscle reflexes that can alter trunk muscle recruitment patterns and co-activation (Hodges, 2001; Lephart et al., 1997). Alternation of trunk muscle firing patterns and co-activation can decrease trunk stiffness that assists in stabilizing the spine. (Granata & Orishimo, 2001; van Dieen, Kingma, & van der Bug, 2003). Thus, back proprioceptive deficits combined with weakness of the back muscles may result in instability of the lumbar spine in golfers with a history of LBP.

It was suggested that the number of mechanoreceptors in the lumbar facet capsules is small and that each receptor may be responsible for a relatively large receptive field (McLain & Pickar, 1998). Damage to a small area may denervate the facet and have important implications for long-term spinal joint function (McLain & Pickar, 1998). Among golfers with LBP, Sugaya et al. observed that right-handed golfers with low back symptoms exhibited a higher rate of right side vertebral and facet joint degeneration than non-golfing controls (Sugaya et al., 1999). Based on the findings of McLain et al. and Sugaya et al., it was hypothesized in the current study that golfers with LBP may have back proprioceptive deficits in more than one movement plane (McLain & Pickar, 1998; Sugaya et al., 1999). Spinal repositioning errors in trunk right rotation and right side bending in the LBP group were not significantly worse than the control group after corrections were used to control for an inflated Type I error. However, right-handed golfers with LBP may still have proprioception deficits in those two directions of trunk movement based on the probability values revealed in this study. It can be speculated that left-handed golfers with LBP may have similar back proprioception deficits in their trunk rotation and side bending to the left.

4. Postural Stability

Postural stability can be disturbed in the presence of impairment in strength, coordination, and/or effective coupling of muscles in the lumbar and pelvic area (Luoto et al., 1998). Reduced postural stability may also indicate the impairment of the neuromuscular feedback loops at different levels of motor activation within the central nervous system (Ebenbichler et al., 2001). Previous research has demonstrated that individuals with LBP presented greater postural sway during standing balance tests, especially with increased task complexity (Luoto et al., 1998; Mientjes & Frank, 1999). Golfers with LBP were generally weaker in the trunk and hip in the current study and demonstrated back proprioceptive deficits. However, these golfers did not show postural instability when compared to the group without LBP. All subjects in this study were proficient golfers who may be more physically fit than those individuals who are not athletes. The balance tests conducted in the current study may not provide sufficient challenge for them to differentiate the capability of postural control between the two groups. In addition, individuals who showed poor postural control in the studies of Luoto et al. and Mientjes et al. were symptomatic at the time of testing (Luoto et al., 1998; Mientjes & Frank, 1999). In the study of Luoto et al., three groups of subjects were tested (Luoto et al., 1998). They were individuals without LBP, with moderate LBP, and individuals with severe LBP. Only individuals with sever LBP demonstrated significantly worse postural control than the healthy group. Golfers in this study were tested when they were pain free. This may be another reason why no differences presented in the postural stability between the two groups. Pain plays a significant role in disturbing the sensorimotor system for postural control. Having good postural stability may contribute to a golfer's swing and also reduce the chance of getting low back injuries.

C. Biomechanical Analysis of Trunk Motion During the Golf Swing

1. Trunk Kinematics During the Golf Swing

X Factor

A large maximum X factor at the beginning of the downswing increases storage of potential energy for maximum clubhead speed at impact when the potential energy becomes kinetic energy. However, it was hypothesized that if a golfer generates a maximum X factor beyond their physical limitation of trunk rotation, excessive stresses may contribute to ongoing irritation of the spinal structures and lead to the development of low back injury (Lindsay & Horton, 2002). The results of the current study revealed that the LBP group, on average, had less maximum X factor than the group without LBP. The maximum X factor normalized by the maximum right trunk rotation angle in neutral position though, was also not significantly different between golfers with and without LBP. These findings are different from the results in the study of Linday and Horton (Lindsay & Horton, 2002). These authors found that golfers with LBP had maximum X factors similar to the golfers without LBP. They also found that the maximum X factors in the LBP group were beyond their ROM in trunk right rotation and that may contribute to their low back injuries. Based on the different findings between the current study and the study of Linday and Horton (Lindsay & Horton, 2002), whether rotating upper torso beyond physical limitation of trunk rotation during the backswing contributes to lower back injury was not answered. However, golfers with LBP demonstrated significantly less right trunk rotation flexibility compared to the healthy group. Improving flexibility of right trunk rotation for golfers with LBP may decrease the stress to their back structures and may therefore reduce the risk of recurrent low back injuries during the golf swing for them.

Crunch Factor

It was hypothesized that the combination of lumbar lateral bending and spinal rotation velocities during the downswing would contribute to lumbar degeneration and injury as these fast movements can generate a large lateral bending moment and large rotational moment in the lumbar spine (Morgan et al., 1997). Thus, it was proposed that a golfer with greater maximum crunch factor would have an increased risk of low back injury (Morgan et al., 1997). However, the results of this study revealed that the LBP group tended to have less maximum crunch factor during the golf swing. Similar findings were also reported by Lindsay and Horton (Lindsay & Horton, 2002). The authors calculated crunch factor using the product of trunk lateral bending and trunk rotation velocity and found a smaller maximum trunk crunch factor in the LBP group when compared to the golfers without LBP. The results reveal that regardless of the chosen anatomical location for swing mechanics, there is a trend that golfers with LBP demonstrated less maximum crunch factor. Plausible reasons for these results are that the LBP group may have intentionally avoided the swing pattern that creates great crunch factor to protect back from re-injury; or the crunch factor may not be a variable for assessing low back injury risk during the golf swing like Lindsay and Horton (Lindsay & Horton, 2002) suggested.

Spinal Rotation Velocity

Hosea and associates reported that rapid spinal rotation velocity could produce considerable amounts of spinal load during the golf swing and result in the development of low back injuries (Hosea et al., 1990). Since low back injuries in golfers are thought to be caused by the forces that are associated with lumbar movements, lumbar spinal rotation velocities were calculated for both groups of golfers in this study. Similar maximum lumbar spinal rotation velocities during the golf swing were found between golfers with and without LBP. Moreover, rotational velocity based on the movement of whole trunk was also examined. No significant difference in trunk rotational velocity was observed. Lindsay and Horton reported similar findings that there were not significant differences in trunk rotational velocities between golfers with and without LBP (Lindsay & Horton, 2002). These results may indicate that although rapid trunk rotation during the golf swing produces large spinal loads, it may not be the sole contributor to low back injuries. Rapid trunk rotation during the golf swing combined with physical limitations, however, may play a role in this golf specific injury.

Incline Factor

Forward trunk flexion and lateral bending angles may lengthen the lever arm of the upper torso's center of mass and may increase the shear forces to the lumbar spine. It was also suggested that the inappropriate combination of trunk flexion and lateral bending may limit the amount of trunk rotation during the golf swing causing greater shear forces in the spine (Geisler, 2001). Lindsay and Horton demonstrated no significant differences in maximum trunk flexion and right side bending angles between golfers with and without LBP during the golf swing (Lindsay & Horton, 2002). Thus, the incline factor (calculated as the instantaneous product of trunk flexion angle and lateral bending angle) was hypothesized to have a significant impact on low back injuries by increasing spinal loads during the golf swing in this study. However, the maximum incline factors of both golfers with and without LBP were not significantly different in this study.

Reverse C Position

Trunk hyperextension at the end of golf swing has been considered as a risk factor to the low back injuries by increasing spinal forces (Geisler, 2001). However, golfers with LBP did not demonstrate a different trunk extension angle at the end of swing than golfers without LBP in the current study. Lindsay and Horton observed that golfers with LBP demonstrated less maximum trunk extension angle at the end of swing than the golfers without LBP (Lindsay & Horton, 2002). Lindsay and Horton described this finding as a potential protective mechanism adopted by injured golfers to prevent LBP as the golfers in their study were tested with existing pain (Lindsay & Horton, 2002). In the current study, the LBP golfers were tested with no current musculoskeletal or neurological symptoms. They may have avoided hyperextending their back to prevent the occurrence of LBP. Likewise, the trunk extension angle at the end of golf swing may not have been a major contributor to their low back injuries.

2. <u>Lumbar Spinal Kinetics During the Golf Swing</u>

Hosea et al. and Lim et al. estimated spinal loads during the golf swing using different mathematical models (optimization vs. EMG-assisted optimization model) (Hosea et al., 1990; Lim & Chow, 2000). Hosea et al. reported that large compressive forces generated during the golf swing (8 body weights) may injure the intevertebral disc and the pars interarticularis (Hosea et al., 1990). Lim and Chow, on the other hand, reported that maximum compressive forces generated during the golf swing was about seven body weight (Lim & Chow, 2000). Lim and Chow considered the accumulated stress and not the magnitude of the load as the primary etiology of low back injuries (Lim & Chow, 2000). Unlike the studies of Hosea et al. and Lim et al., this study used a bottom-up dynamic 3D linked segment model to examine the differences in the spinal loads at the L5/S1 level during the golf swing between golfers with and without LBP. Trunk muscle activities (electromyographic activities) were not a variable in the calculation of the spinal loads. The LBP group demonstrated similar maximum spinal forces and moments at L5/S1 during the golf swing compared to the healthy group. The LBP group also demonstrated similar spinal forces and moments at the seven swing points. However, the standard deviation of each force and moment was high among each group of golfers. Although the spinal loads at L5/S1 varied among golfers and did not show significant differences between the two groups, the results of this study provide valuable information about an overall pattern of each spinal force and moment in each group of golfers. The results may also imply that golf swing can generate considerable amount of spinal forces and moments similar for both groups of golfers.

D. Summary

The results of this study demonstrated that golfers with LBP had decreased trunk and hip strength and a strength imbalance between the trunk extensors and flexors.

The LBP group also had limited hamstring and trunk rotation flexibility toward non-lead side. These strength and flexibility deficits were coupled with trunk proprioception deficits, especially trunk flexion postural awareness. Golfers with LBP may suffer low back injuries due to the inappropriate combination of physical characteristics that prevents dissipation of the tremendous spinal forces and moments generated by the golf swing over time.

Core stability has been considered to be important for lumbopelvic control with varied rationales (Hodges, 2003). It represents an ability of the neuromuscular system to control and protect the spine from injury or re-injury (Hodges, 2003). If trunk and hip muscle strength do not meet the demands of control and the coordination and control of the trunk muscles also have deficits, the stability of the lumbopelvic region (the core of the body) would diminish and lead to spinal instability (Hodges, 2003; McGill, 2002; Richardson, Jull, Hodges et al., 1999; Vleeming, .Mooney, Dorman et al., 1997). Spinal instability may contribute to repetitive injuries and a progressive decline of core stability (Lephart & Fu, 2002; Richardson et al., 1999). The LBP golfers in the current study demonstrated decreased trunk and hip strength and decreased back proprioception. This may reveal that these golfers presented the clinical definition of having poor core stability. Therefore, improving core stability is important for golfers, especially during the golf swing which can generate considerable amounts of load to the spine. This loading requires a strong core to balance these forces.

Furthermore, limitations in right trunk rotation may result in the inability to generate high amounts of potential energy at the top of the back swing by maximizing the coiling effect of rotating the shoulders relative to the pelvis, or the X factor. This limitation usually results in a swing that produces less than desired hitting distance. As a result, the golfer may try to rotate upper body beyond their physical limitation of trunk rotation. Occasionally this would place excessive stress on the back resulting in soft tissue injury from a single swing.

Knowledge about the differences in swing mechanics and physical characteristics that may contribute to the low back injuries is still vague. Low back injury in golfers may result from a single swing or develop gradually from chronic loading due to swinging a club with sub-optimal physical fitness. Both can contribute to permanent disability. It is unknown if the deficiency found in the LBP group in the current study contributed to the back injury or are the result of the injury. Regardless, clinicians may still be able to use this information for designing appropriate back-specific exercise programs for golfers to help prevent or rehabilitate low back injuries. To date, many exercise programs have been designed for low back injury prevention and rehabilitation by clinicians and golf teaching professionals without a scientific basis for its content. Such programs may not be effective if they do not address the physical deficiencies of the golfer revealed in the current study. These results provide comprehensive information related to the physical characteristics of golfers with low back injury and as such furthers the knowledge in this area.

E. Limitation of the study

Spinal forces and moments during the golf swing were calculated using a bottom-up dynamic 3D LSM in this study. The benefit of using this model is that the LSM can be a powerful tool for routine examination of physical demands of an activity. The complex EMG-assisted models, however, may provide more insight as to how injury occurs with various loads generated by trunk muscle contractions (McGill, 2002). Hosea et al. and Lim et al. have tried to estimate spinal loads during the golf swing by adding the forces that may be produced by trunk muscle activities (Hosea et al., 1990; Lim & Chow, 2000). The limitation is that golf swing involves rapid trunk movements. Trunk muscles do not contract at a static or controlled speed during the golf swing and the length-tension relationship of the trunk muscles is not easy to control. Without satisfying these conditions, it will not be easy to provide good prediction of muscle forces based on the current models.

Another limitation of this study was that although golfers with LBP demonstrated certain level of deficits in back proprioception, the intraclass correlation coefficients of the testing protocol were relative low. The protocol for the current study was conducted for assessing back proprioception because the standard error of measurements of the testing protocol was low and the reliability of the testing equipment was high. It was expected that the results of this study could show clinical differences in back proprioception between golfers with and without LBP. These clinical findings could provide some information for clinicians to design injury prevention and rehabilitation programs for golfers. This study has shown proprioception deficits in golfers with LBP. Further research should refine the testing protocol for assessing back proprioception in order to get more accurate measurements.

F. Future Research

The golf population is continuing to grow worldwide. Injury prevention and rehabilitation are critical to enjoyment of the game. Deficits in physical characteristics have been observed in golfers with and without a history of LBP. Future research should focus on designing a comprehensive back injury prevention and rehabilitation program specifically for golfers based on the findings of this study which also examine the effects of the designed exercises. This training program should include the following exercises for golfers. Trunk and hip muscle strengthening exercises and back proprioception training are required for improving core stability. While designing proprioception exercises for golfers, clinicians should be aware that right-handed golfers may have proprioception deficits in right trunk rotation and right trunk side bending and left-handed golfers may have proprioception deficits in left trunk rotation and left trunk side bending in addition to deficits in trunk flexion. Stretching exercises for increasing the flexibility of trunk rotation and hamstrings are also crucial. They can provide greater ROM for storing more potential energy required during the downswing without creating excessive stress to the back. Moreover, balance training is also important for improving postural stability during the golf swing. Balance training can be conducted on uneven surfaces to increase the task difficulty. In an attempt to determine an optimal physical fitness level for golfers, several recommendations are made in Table 5.1.

Table 5.1 Recommendations for an optimal physical fitness for golfers

Physical characteristics			*Values*	*Unit*
Strength	*Trunk (60 deg/sec)*	*Extension*	*> 360*	*%BW*
		Flexion	*> 200*	*%BW*
		Right rotation	*> 140*	*%BW*
		Left rotation	*> 140*	*%BW*
	Bilateral hips (isometric)	*Abduction*	*> 155*	*%BW*
		Adduction	*> 156*	*%BW*
		Extension	*> 320*	*%BW*
		Flexion	*> 85*	*%BW*
Flexibility	*Trunk*	*right rotation*	*> 47*	*deg.*
	Hamstring (Active knee extension test)	*Knee flexion angle*	*< 16*	*deg.*
Back proprioception	*Spinal repositioning error*	*in each direction*	*< 2*	*deg.*
Postural stability	*Balance training on uneven surface*			

%BW = % body weight

APPENDIX A

FLYER

SUBJECTS NEEDED FOR A GOLF RESEARCH STUDY

What: Study comparing golf swing, strength, flexibility, back position sense, and balance between golfers with low back pain and healthy golfers

Where: Neuromuscular Research Laboratory
University of Pittsburgh
3200 South Water Street

Who: Healthy golfers and golfers with a history of low back pain within the past two years

Ages between 18-65 years old with an established USGA handicap better than 20

If interested, please contact:

Yung-Shen Tsai, MA, PT
yut4@pitt.edu
(412) 432-3800

APPENDIX B
OSWESTRY QUESTIONNAIRE

Subject: _______________ **Date: _______________**

This Questionnaire is designed to enable us to understand how much your Low Back Pain had affected your ability to manage your everyday activities. Please answer each section by marking in each section the ONE BOX that most applies to you. We realize that you may feel that

Section 1 - Pain Intensity

○ I can tolerate the pain I have without having to use pain medication
○ The pain is bad, but I can manage without having to take pain medication
○ Pain medication provides me with complete relief from pain
○ Pain medication provides me with moderate relief from pain
○ Pain medication provides me with little relief from pain
○ Pain medication has no effect on my pain

Section 2 – Personal Care (washing, dressing etc.)

○ I can take care of myself normally without causing increased pain
○ I can take care of myself normally, but it increases my pain
○ It is painful to take care of myself, and I am slow and careful
○ I need help, but I am able to manage most of my personal care
○ I need help every day in most aspects of my care
○ I do not get dressed, I wash with difficulty, and I stay in bed

Section 3 – Lifting

○ I can lift heavy weights without increased pain
○ I can lift heavy weights, but it causes increased pain
○ Pain prevents me from lifting heavy weights off the floor, but I can manage if the weights are conveniently positioned (e.g., on a table)
○ Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned
○ I can lift only very light weights
○ I cannot lift or carry anything at all

Section 4 – Walking

○ Pain does not prevent me from walking any distance
○ Pain prevents me from walking more than 1 mile (1 mile = 1.6 km).
○ Pain prevents me from walking more than 1/2 mile
○ Pain prevents me from walking more than 1/4 mile
○ I can walk only with crutches or a cane
○ I am in bed most of the time and have to crawl to the toilet

Section 5 – Sitting

○ I can sit in any chair as long as I like
○ I can only sit in my favorite chair as long as I like
○ Pain prevents me from sitting for more than 1 hour
○ Pain prevents me from sitting for more than 1/2 hour
○ Pain prevents me from sitting for more than 10 minutes
○ Pain prevents me from sitting at all

Section 6 – Standing

○ I can stand as long as I want without increased pain
○ I can stand as long as I want, but it increases my pain
○ Pain prevents me from standing for more than 1 hour
○ Pain prevents me from standing for more than 1/2 hour
○ Pain prevents me from standing for more than 10 minutes
○ Pain prevents me from standing at all.

Section 7 – Sleeping

○ Pain does not prevent me from sleeping well
○ I can sleep well only by using pain medication
○ Even when I take medication, I sleep less than 6 hours
○ Even when I take medication, I sleep less than 4 hours
○ Even when I take medication, I sleep less than 2 hours
○ Pain prevents me from sleeping at all

Section 8 – Social Life

○ My social life is normal and does not increase my pain
○ My social life is normal, but it increases my level of pain
○ Pain prevents me from participating in more energetic activities (e.g., sports, dancing)
○ Pain prevents me form going out very often
○ Pain has restricted my social life to my home
○ I have hardly any social life because of my pain

Section 9 – Traveling

○ I can travel anywhere without increased pain
○ I can travel anywhere, but it increases my pain
○ My pain restricts my travel over 2 hours
○ My pain restricts my travel over 1 hour
○ My pain restricts my travel to short necessary journeys under 1/2 hour
○ My pain prevents all travel except for visits to the physician / therapist or hospital

Section 10 – Employment/Homemaking

○ My normal homemaking/job activities do not cause pain
○ My normal homemaking/job activities increase my pain, but I can still perform all that is required of me
○ I can perform most of my homemaking/job duties, but pain prevents me from performing more physically stressful activities (e.g., lifting, vacuuming)
○ Pain prevents me from doing anything but light duties
○ Pain prevents me from doing even light duties
○ Pain prevents me from performing any job or homemaking chores

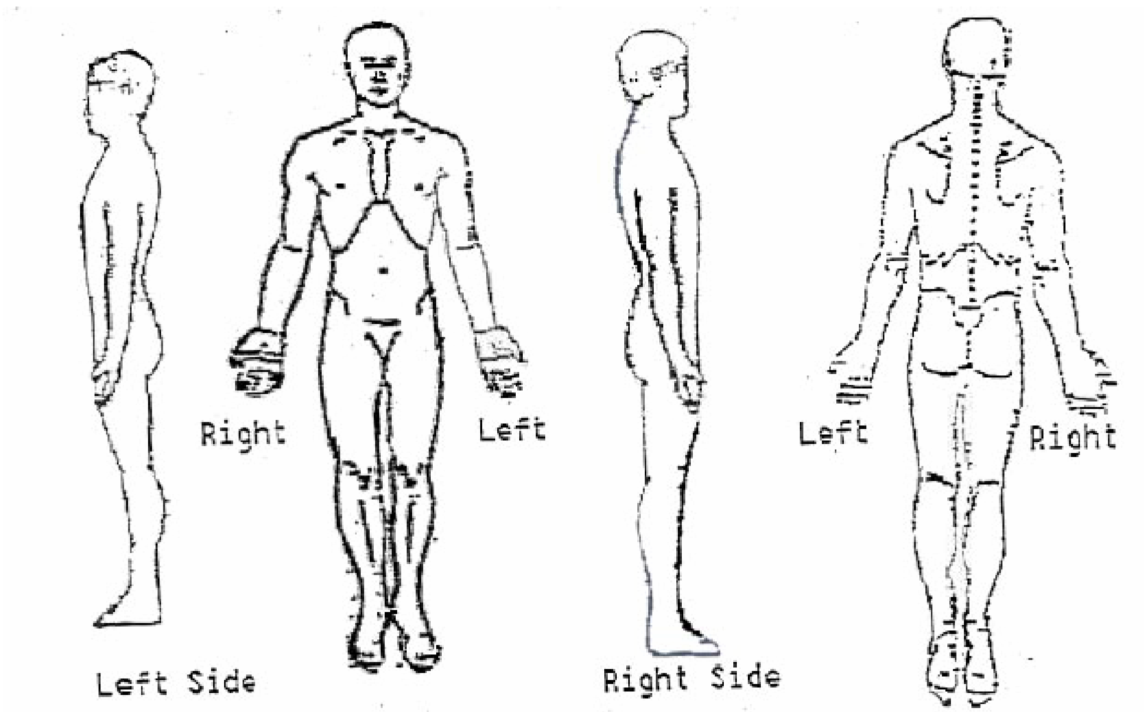
APPENDIX C

PAIN SCALE AND PAIN DIAGRAM

Subject: _______ **Date: ________**

Please use the diagram below to indicate where you felt symptoms during the worst episode within the past two years. Use the following key to indicate the different types of symptoms.

Key: Pins and Needles = ooooo, Stabbing = /////, Burning = xxxxx, Deep ache = zzzzz



Please use the scale below to rate your worst pain during the past two years.

0 = NO PAIN 10 = EXTREMELY INTENSE

O **0** O **1** O **2** O **3** O **4** O **5** O **6** O **7** O **8** O **9** O **10**

APPENDIX D

BASIC NEUROLOGICAL EXAM TO SCREEN FOR NERVE ROOT COMPROMISE

Subject:_______ Date: ________

1. Knee Jerk Reflex (Patellar Reflex) – L4 nerve

	Normal	Hypo	Hyper
Right			
Left			

2. Straight Leg Raising (SLR) – Sciatic nerve

	ROM (deg)
Right	
Left	

3. Toe & Heel Walking – S1 (toe) & L5 (heel) nerve

	Toe Walking	Toe Walking
Right		
Left		

COMMENTS:

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