SPINAL MECHANICS DURING DROP LANDING: EFFECTS OF GENDER, FATIGUE AND LANDING TECHNIQUE

By

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A DISSERTATION PRESENTED TO THE GRADUATE SCHOOL
OF THE UNIVERSITY OF FLORIDA IN PARTIAL FULFILLMENT
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To my parents

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By

Soo-An Park

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Major: Health and Human Performance

Objective: To investigate the kinematics of the spinal column and the kinetics of lumbosacral (L/S) and cervicothoracic (C/T) junctions during the drop landing, and to evaluate the effects of gender, landing technique, and fatigue.

Methods: Thirteen male and 13 female healthy young volunteers were tested. To track the kinematics of different spinal regions, surface markers were placed on skin over selected spinous processes. Data were collected using a 3-D motion capture system and a forceplate. The subject performed 3 drop landings using his/her own landing technique (NL) and 3 soft landings with instruction (SL; SL1). During each trial, the subject descended from a 50-cm height platform and landed on a forceplate with the left foot at the center of forceplate. After completing isokinetic knee flexion/extension exercises and a 30-minute run on a motorized treadmill, the subject performed 3 more soft landings in a knee joint muscle fatigue state (SL2). Kinematic variables included touchdown angle and initial extension range of motion of different spinal regions. Kinetic variables included joint resultants at L/S and C/T junctions computed using an inverse dynamics approach. Multivariate analyses of variance (MANOVA) and follow-up univariate analyses of variance (ANOVA) were used to examine the effects of gender, landing technique, and fatigue on different kinematic and kinetic variables.

Results: Females exhibited a significantly extended thoracolumbar landing posture and greater thoracic motion than males in all 3 experimental conditions. Thoracic and lower cervical extension motions increased and most joint resultants decreased significantly when going from NL to SL. Posterior shear force in males and anterior shear force in females were significantly greater than their counterparts at C/T junction during NL. Females exhibited significantly increased joint resultants from SL1 to SL2, while males did not.

Conclusion: The spinal column is more actively involved in energy absorption during drop landings in males, and the thoracolumbar region could be more loaded by hyperextension during soft landing in females comparing to males. Repeated drop landings may cause injuries to the cervical spine by different mechanisms in each gender. For females, soft landings under fatigue condition can be a risk factor of spinal injury.

CHAPTER 1
INTRODUCTION

Spinal function in vivo has been generally determined by radiographs, showing the kinematics of each spinal segment and spinal region (Wong et al., 2006) or by inclinometer measurements (Ng et al., 2001). Most previous investigations about spinal kinematics were based on postures of the spinal column in static conditions and static spinal kinematics. Active spinal motion can be identified as the end range of motion (ROM) achieved by the subjects, and the passive motion is the end ROM obtained by applying external forces to a fully motioned spine. Both active and passive ROM of the spine have been used to interpret the functional status of the spine in clinical and laboratory studies (Dvorak et al., 1988).

Several researches have evaluated the kinematics of the spine in dynamic situations like walking (Callaghan et al., 1999; Crosbie et al., 1997) and running (Schache et al., 2002). Although results from both walking and running analyses may have clinical implications, these two locomotive tasks may not be the most adequate tasks to reveal spinal function in dynamic situation.

The age of patients who develop spinal degeneration and undergo spinal surgeries is getting younger and this population is getting larger (Kjaer et al., 2005). The prevalence of spinal pain and degeneration is higher in active individuals (Bono, 2004). The biomechanics of the spine in a static condition are quite different from those in a dynamic condition. Demands for developing a functional evaluation of the spine *in vivo* have increased, because conventional radiographic study of spinal ROM is not good enough to estimate the spinal function of every subject. Also with the development of the surgical technique of spinal arthroplasty, the restoration of the original function of each patient has become a primary goal of the surgery.

Therefore, normal spinal mechanics *in vivo* in various tasks and the mechanical factors contributing to spinal pain and degeneration are useful information to medical practitioners.

It has been suspected that most disabling and chronic back pain arises from intervertebral disc degeneration. With disc degeneration, biologic as well as biomechanical changes follow. Disc degeneration occurs most commonly in the third to fifth decades of life. Aging causes definite changes in the morphology and composition of spinal tissues that are mostly unrelated to pain. However, many studies reported that aging weakens the intervertebral disc tissues due to decreased cell number, high apoptosis rate, and the different response to biologic environments. Impaired function of the intervertebral discs may make people more vulnerable to mechanical injuries, which can initiate further structural and symptomatic disc degeneration.

Nonphysiologic loading to spinal structures can contribute to intervertebral disc degeneration. With the increased loading to the trunk, spinal shrinkage was found double to the unloaded condition (Fowler et al., 1994). A flexion posture significantly increases extensor muscle activity when compared with a standing neutral posture (Arjmand & Shirazi-Adl, 2006). Forward leaning of the trunk causes the vertebrae in anterior translation, and disc loads and stresses were significantly increased most markedly at the L5/S1 level (Harrison et al., 2005). When the torso is fully flexed during repetitive lifting tasks, fatigue failure of spinal tissues can occur rapidly (Wrigley et al., 2005). On the other hand, abnormally low loading causes atrophy in muscle, cartilage and bone, leaving them less able to resist high loads (White & Panjabi, 1990).

Abnormalities in the lower extremities can affect spinal mechanics. For example, patients who have leg length discrepancy due to lower extremity disorders demonstrate different spinal kinematics during gait when compared with normal subjects suggesting greater risk of

developing spinal disorders (Kakushima et al., 2003). However, a walking task may not be sensitive enough to detect the effects of altered mechanics in the lower extremity upon spinal mechanics.

Gender differences in spinal kinematics during trunk posturing hinted that there may be different mechanical causes of spinal degeneration in different genders. During a prolonged sitting, males exhibit more flexion of lumbar spine and trunk than females. Males and females may be exposed to different loading patterns during certain prolonged postures and can develop different injuries or degeneration mechanisms of the spine (Dunk & Callaghan, 2005).

Various *in vivo* and *in vitro* biomechanical techniques have been developed to investigate spinal mechanics, but they all have limitations. Physical properties of the human spine may be obtained from studies of living subjects, whole cadavers, isolated whole cadaveric spines, and isolated spinal segments. A living subject provides realistic but less accurate measurements. An isolated spinal segment lacks muscles, but can provide accurate data and allows the possibility of studying the effects due to trauma and surgical stabilizations (White & Panjabi, 1990).

The force applied to the spine depends on body weight (Rodacki et al., 2005), external loads (Lawrence et al., 2005), and internal muscle forces (Arjmand & Shirazi-Adl, 2006) which can be varied during dynamic activities (Chow et al., 2003; Tully et al., 2005). Direct measurement of spinal compression force pioneered by Nachemson (1966; 1964), obtained by inserting a pressure needle into the lumbar intervertebral discs of living subjects, has been benchmarked and compared to many other studies that measured forces applied to the spine *in vivo* (Ledet et al., 2005; Sato et al., 1999). However, most researchers who have investigated spinal mechanics utilized indirect techniques on living subjects and cadaveric spinal segments because of the invasiveness and limited localization of direct measurements.

Normal spinal mechanics during activities of living subjects and each spinal segment as functional spinal unit (FSU) and spinal column are usually studied using indirect measurements. Spinal mechanics during various movements – walking, forward/backward bending, sit-to-stand/stand-to-sit, etc. – have been investigated using image-based motion analysis systems. Using surface electromyography (EMG) techniques, spinal muscle activity during different spinal movements has been investigated. McGill (1992) estimated the moments generated by trunk muscles using an EMG-driven musculoskeletal model during trunk posturing movements. By applying inverse dynamic techniques to a rigid segment model, joint resultants at different lower extremity joints and spinal motion segments can also be calculated. However, many spinal vertebrae included in one trunk segment may mimic the joint resultant values at the trunk segment, and the kinematics of the spinal region *in vivo* could not be accessed in details (Khoo et al., 1995).

Using forceplate and kinematic data, inverse dynamic techniques are commonly used to calculate joint resultants at lower extremity joints during various activities (Kernozek et al., 2005). Jumping and landing were commonly adopted for measuring joint resultants at lower extremity joints simulating active and vigorous movements. A recent study completed by the author indicated that limiting trunk movements caused changes in the mechanics of lower extremity joints during drop landings (Park et al., 2006b). This finding suggests that body segments proximal to the hip joints could be involved in regulating the force transmitted to the lower extremity joints during the landing phase. Conversely, different mechanical configurations of lower extremity joints may affect spinal mechanics during landing and the findings from spinal mechanics for each specific configuration may provide insights into the spinal function in dynamic situations.

Purpose of the Study

The purpose of this study was to investigate the kinematics of different regions of the spinal column and the kinetics at the lumbosacral (L/S) and cervicothoracic (C/T) junctions during drop landings. Specifically, this study evaluated the effects of gender and landing techniques (self-selected landing and instructed soft landing techniques) on spinal mechanics during drop landing in the first part of this study. In the second part of this study, the effects of gender and muscular fatigue of knee flexor/extensor on spinal mechanics during drop landings were examined.

Significance of the Study

Previous researchers have evaluated the mechanical causes of spinal degeneration *in vivo* and *in vitro* while simulating various activities using direct and indirect measurement techniques. Most activities employed in these biomechanical studies were simple, everyday activities that do not demand much spinal movement. However, the populations who have spinal degeneration and undergo spinal surgeries are getting larger and younger. These individuals want to be physically active and participate in activities that may demand vigorous spinal movements. Drop landings have been widely used to examine coordination and mechanical stress at different joints under dynamic situations. However, most landing studies were confined to the biomechanics of lower extremity joints and very few studies evaluated mechanics of upper body movements during landings.

In addition to lower extremity joints, this study attempted to identify mechanical characteristics of different spinal regions during drop landings. The results might provide insight into the effects of gender, landing technique and fatigue of knee joint muscles on spinal mechanics during drop landings.

Hypotheses

In the absence of extensive pilot data and relevant data reported in the literature, the following questions were raised and null hypotheses were tested to address the aims of this study:

- To determine the effects of gender and landing technique on spinal column kinematics and loads to the lumbosacral (L/S) and cervicothoracic (C/T) junctions during drop landings.

  Q1: Would there be significant differences in spinal column kinematics and loads to the L/S and C/T junctions during drop landings between males and females regardless of landing technique?

  1a. There would be no significant differences in motion characteristics of the spinal column between males and females.

  1b. There would be no significant differences in peak resultant forces and moments transmitted to the L/S and C/T junctions between males and females.

  Q2: Would the self-selected and soft landing techniques cause significant differences in spinal column kinematics and loads to the L/S and C/T junctions during drop landings?

  2a. The conditions for landing technique would not cause significant differences in motion characteristics of the spinal column.

  2b. The conditions for landing technique would not cause significant differences in peak resultant forces and moments transmitted to the L/S and C/T junctions.

- To determine the effects of gender and fatigue of knee joint muscles on spinal column kinematics and loads to the L/S and C/T junctions during drop landings using a soft landing technique.

Q3: Would there be significant differences in spinal column kinematics and loads to the L/S and C/T junctions during drop landings using a soft landing technique between males and females regardless of fatigue condition?

3a. There would be no significant differences in motion characteristics of the spinal column between males and females.

3b. There would be no significant differences in peak resultant forces and moments transmitted to the L/S and C/T junctions between males and females.

Q4: Would fatigue of knee joint muscles cause significant differences in spinal column kinematics and loads to the L/S and C/T junctions during drop landings using a soft landing technique?

4a. Fatigue of knee joint muscles would not cause differences in motion characteristics of the spinal column.

4b. Fatigue of knee joint muscles would not cause differences in peak resultant forces and moments transmitted to the L/S and C/T junctions.

Limitations

- Measurement errors of forceplate and digital video cameras are always present but they are considered acceptable within the specifications of the manufacturers.
- Marker placement was controlled cautiously to minimize errors.
- Sagittal spinal kinematics relative to the adjacent spinal region based on spinal marker locations would have some errors due to skin movement, but will be considered acceptable.

- The center of L/S junction was assumed to be located at the midpoint between both posterior iliac crest markers at L5/S1 level. The center of the C/T junction was assumed to be located at the midpoint between the two acromial process markers.
- Mechanical characteristics of the lower extremities were assumed to be symmetrical and only the data collected from the left leg were used to calculate joint resultants at the L/S and C/T junctions.

CHAPTER 2
MATERIALS AND METHODS

Subjects

Thirteen male (age: 21.4 ± 1.3 yrs, mass: 74.2 ± 10.2 kg, height: 174.8 ± 5.2 cm) and 13 female (age: 21.1 ± 1.3 yrs, mass: 58.6 ± 6.8 kg, height: 165.5 ± 5.3 cm) healthy and physically active individuals participated in this study. They were free from any cardio-respiratory diseases that would prevent them from completing the fatigue procedures and musculoskeletal diseases or injuries that could influence spinal and lower extremity mechanics. Before testing, each subject carefully read and signed a written informed consent approved by the Institutional Review Board of the University of Florida (Appendix A).

Sample Size Justification

To simplify the calculation of sample size determination, data of a dependent variable (peak extensor moment at the lumbosacral junction) from a pilot study were used to set up a 2 × 2 [gender × landing technique: normal landing and soft landing] ANOVA with repeated measures on the last factor. Based on the preliminary data collected from one male and one female, the means of male and female data were 2.74 and 2.58 $N \cdot m \cdot kg^{-1} \cdot BH^{-1}$, respectively, and those for NL and SL were 3.05 and 2.27 $N \cdot m \cdot kg^{-1} \cdot BH^{-1}$, respectively (Note. BH: body height). There was a 39% difference between male and female, and a 26% difference between normal and soft landing conditions. The mean standard deviation value was 0.3 (Table B-1 in Appendix B).

Based on these values, the current study was designed to detect at least 12% changes in peak extensor moment in between-subject group and 10% difference in within-subjects groups with alpha = 0.05 and beta = 0.2 (80% power). A Geisser-Greenhouse correction report indicated that power values were over 80% for all the terms with a sample size of n=13 (Table 2-1, Figure

2-1). Therefore, at least 12 subjects were required for each within-subject group to determine the effects of the treatment. All the calculations were performed using PASS 2005 (Number Cruncher Statistical Systems, Kaysville, Utah).

Experimental Setup

A forceplate (Type 4060-10, Bertec Corporation, Columbus, OH) operating at 1,200 Hz was set up at the center of the Biomechanics Research Laboratory (Figure 2-2). A 50-cm height platform was placed behind and slightly to the right of the forceplate. Seven Hawk digital cameras (Motion Analysis Corp., Santa Rosa, CA) were stationed around the forceplate to collect kinematic data and were 3-4 m from the landing area. Kinematic data were captured at a frequency of 100 Hz. The system was calibrated prior to each testing session according to the procedures specified by the manufacturer.

Testing Protocol

To expose the lower extremity and the back of the trunk, subjects were asked to wear only short pants (both males and females) and sports bra (for females only) (Figure 2-3). They wore their own sports shoes during testing. Upon completing the measurements of body weight and height, each subject jogged on a treadmill and stretched with self-selected exercises for 10 minutes as a warm-up (Figure 2-4).

Reflective markers (1.0 cm in diameter) were placed on the left second metatarsal head, dorsal navicular surface, heel, lateral malleolus, lower-shank, mid-shank, lateral tibial epicondyle, lower-thigh, mid-thigh, greater trochanter, anterior superior iliac spines, 2nd spinous process on median sacral crest of sacrum to track the locations of the left lower extremity and pelvis (Kadaba et al., 1990; Kadaba et al., 1989). Another set of reflective markers was applied over the subject's spinous processes for measuring the kinematics of the spinal column (C4, C6, T1, T3, T6, T10, T12, L2, L4, both posterior iliac crests at L5/S1 level) (Figure 2-5).

After marker placement, drop landing practice trials were provided to ensure consistent landing during experimental trials. In each trial, the subject descended from a 50 cm-height platform and landed on a forceplate with his/her left foot at the center of the forceplate and right foot on a wooden platform of the same height.

Pre-Fatigue Landing Trials

After practice drop landings, the subject completed 3 trials of drop landing using his/her own landing technique (normal landing: NL). The subject was then instructed on how to perform a drop landing using the soft landing technique (soft landing: SL). He/She was instructed to try soft landing by touching the balls of feet on the ground at initial impact, delaying heel contact, and using more knee flexion after the landing. Three trials of soft landing were collected (SL1). The averages over the three trials for each landing technique were used in subsequent analyses. In each trial, kinematic and GRF data were sampled for 4 s. If the subject did not maintain balance after landing, that trial was discarded and repeated.

Fatigue Procedure

The subject was asked to sit on the chair of a KinCom dynamometer (Chattanooga Group, Inc., Hixson, TN) and perform 30 repetitions of isokinetic reciprocal knee flexion/extension with full ROM and maximal effort at 60°/s to induce muscular fatigue of knee joint flexor/extensor. The exercise was repeated at the speed of 180°/s to measure the isokinetic strength for the purpose of quantifying the fatigue level. After the isokinetic exercises, the subject ran on a motorized treadmill at 4-6 mph for 30 minutes. The running intensity was lower than the typical daily exercise for developing and maintaining fitness (Fletcher et al., 2001; Pollock et al., 1998). Also, the speed and duration of running used in this study was known not to elicit cardio-respiratory fatigue (Hardin et al., 2004). If the subject became exhausted before the end of the 30

minutes run, the running speed was reduced to 4.0 mph so that he/she could complete the run. Isokinetic knee flexion/extension exercises were repeated immediately after running to quantify the fatigue state of knee joint muscles.

Post-Fatigue Landing Trials

Immediately after the fatigue procedure, the subject performed 3 trials of drop landing using the soft landing technique from the same platform (soft landing performed under fatigue condition: SL2). The averages over the 3 trials were used in subsequent analyses.

Data Reduction

Kinematic data were processed using EVaRT 4.6 software (Motion Analysis Corp., Santa Rosa, CA). The animation of reflective markers in each trial was examined qualitatively by the investigator. Positional data were smoothed using a Butterworth low pass filter with a cutoff frequency of 10 Hz. Three-dimensional kinematic and kinetic data for the left ankle, knee, and hip joints, pelvis and trunk, and the kinematic data of spinal column markers were calculated using Kintrak 6.2 software (Motion Analysis Corp., Santa Rosa, CA). Locations of spinal column markers were used to define 5 spinal regions: lower cervical (LC), thoracic (TH), thoracolumbar (TL), lumbar (L) and sacral (S) regions (Figure 2-5).

Joint resultants at the left knee and hip joints and L/S and C/T junctions were computed based on the kinematic and forceplate data using an inverse dynamics approach. Assuming symmetry in lower extremity mechanics, mechanical characteristics of the right leg were the same at the left leg for the purpose of computing joint resultants at the L/S and C/T junctions. To minimize the variation due to individual differences in physique, force variables were normalized to the subject's body mass, and moment variables were normalized to body mass and body height.

Muscle fatigue was quantified by the fatigue index. Fatigue index was determined by the decline in peak torque in 30 repetitions, and calculated by the following formula to yield a percent decrease for each isokinetic torque value:

Fatigue index = 100 – [(last 5 repetitions/highest consecutive 5 repetitions) × 100]

For each subject, the highest consecutive five repetitions were determined by the values attained from the two repetitions immediately prior to, and following, the single highest repetition value. If the single highest repetition value was observed within the first 3 repetitions, the first 5 repetitions were used to calculate the fatigue index (Pincivero et al., 2003).

Fatigue levels of knee joint muscles before/after the fatigue procedure were evaluated with the fatigue indices of knee flexor and extensor muscles. Repeated measures MANOVA revealed a significant main effect of knee joint muscles fatigue for the fatigue index variables ($p=0.001$), but did not reveal any significant main effect of gender ($p=0.528$) and interaction between gender and fatigue level ($p=0.589$) (Table 2-2). The univariate contrast procedures indicated that the fatigue indices of both knee flexors and extensors increased significantly by the fatigue procedure (Table 2-3). Only those subjects who demonstrated increased fatigue indices in both knee flexors and extensors (12 males and 10 females) were included in subsequent analyses (SL1 vls. SL2).

The landing phase was defined using the critical instants identified from the kinematic and GRF data (Figure 2-6), and the critical instants are as follows:

- The instant when the vertical ground reaction force (VGRF) starts to increase, the initial touchdown (the beginning of landing phase), was identified from GRF data.
- Maximal knee joint flexion after initial touchdown was identified from the kinematic data (the end of landing phase).

At the completion of data reduction, the dependent variables were divided into five groups:

- Landing variables: peak vertical GRF (PVGRF), time for landing phase ($t_{(LP)}$), knee flexion angle at touchdown ($\theta_{TD(KFA)}$), ROM of knee flexion from touchdown to initial peak of knee flexion ($\theta_{P(KFA)}$), hip flexion angle at touchdown ($\theta_{TD(HFA)}$), ROM of hip flexion from touchdown to initial peak of hip flexion ($\theta_{P(HFA)}$).

- Touchdown angles: lumbar regional angle at touchdown ($\gamma_{TD(L/S)}$), thoracolumbar regional angle at touchdown ($\gamma_{TD(TL/L)}$), thoracic regional angle at touchdown ($\gamma_{TD(TH/TL)}$), lower cervical regional angle at touchdown ($\gamma_{TD(LC/TH)}$).

- Extension ROMs: extension ROM of lumbar region from touchdown to initial peak during landing phase ($\gamma_{P(L/S)}$), extension ROM of thoracolumbar region from touchdown to initial peak during landing phase ($\gamma_{P(TL/L)}$), extension ROM of thoracic region from touchdown to initial peak during landing phase ($\gamma_{P(TH/TL)}$), extension ROM of lower cervical region from touchdown to initial peak during landing phase ($\gamma_{P(LC/TH)}$).

- Kinetic variables at L/S junction: peak axial compressive force [AxF(L/S)], peak anterior shear force [$ShF(L/S)_{ant}$], peak posterior shear force [$ShF(L/S)_{post}$], peak flexor moment [FlxM(L/S)], peak extensor moment [ExtM(L/S)] after touchdown.

- Kinetic variables at C/T junction: peak axial compressive force [AxF(C/T)], peak anterior shear force [$ShF(C/T)_{ant}$], peak posterior shear force [$ShF(C/T)_{post}$], peak flexor moment [FlxM(C/T)], peak extensor moment [ExtM(C/T)] after touchdown.

θ_{KFA} was defined as the angle between the line of shank axis and thigh axis. θ_{HFA} was defined as the angle between the line of thigh axis and pelvis axis. The negative angles of $\theta_{TD(KFA)}$ and $\theta_{TD(HFA)}$ mean the flexions of knee and hip joints at the touchdown. For $\theta_{P(KFA)}$ and $\theta_{P(HFA)}$, absolute values were used.

A regional angle of the spine was defined as the angle between the lines representing a spinal region and its lower adjacent region. A positive touchdown angle indicates the spinal region is in an extended state or extension motion relative to the lower adjacent region and a negative angle indicates the spinal region is in a flexed state or flexion motion (Figure 2-5).

Data Analysis

For the non-fatigued data (NL and SL data), the 5 groups of dependent variables were submitted to five separate 2 × 2 (Gender × Landing type) MANOVA with repeated measures on the last factor. For the soft landing data (SL1 and SL2 data), the 5 groups of dependent variables

were submitted to 5 separate 2 × 2 (Gender × Fatigue level) MANOVA with repeated measures on the last factor. Follow-up univariate analyses were conducted when appropriate. Bonferroni adjustments were used during follow-up testing. A priori alpha level was set at 0.05 for all statistical procedures. All statistical tests were performed using SPSS 13.0 for Windows (SPSS Inc., Chicago, IL).

Table 2-1. Geisser-Greenhouse Correction Detail Report.

Term (levels)	Power	Alpha	F	Lambda	df1|df2	Epsilon	E (Epsilon)	G1
n = 13 N = 26 Means × 1								
Gender (B: 2)	0.8310	0.05	4.26	9.25	1|24	1	1	0
Landing (W: 2)	1	0.05	4.26	43.33	1|24	1	1	0
BW	1	0.05	4.26	43.33	1|24	1	1	0

Table 2-2. MANOVA table for the fatigue indices.

Effect	Roy's Largest Root	F	Hypothesis df	Error df	Sig.(p)	Observed Power (a)
Gender	0.057	0.657	2	23	0.528	0.147
Fatigue*	0.875	10.067	2	23	0.001	0.971
Fatigue × Gender	0.047	0.542	2	23	0.589	0.129

* Significant main effect or interaction (p<0.05)

Table 2-3. Collapsed mean and SD values of fatigue indices before/after the fatigue procedure.

Fatigue indices	Before Mean (SD)	After Mean (SD)	Gender: p	Fatigue: p	Fatigue × Gender: p
Knee extensors					
(%)	14.9 (9.0)	20.2 (11.2)	0.253 (0.2)	0.033 (0.58)*	0.763 (0.06)
M	16.4 (9.3)	22.4 (8.2)			
F	13.4 (8.8)	17.9 (13.4)			
Knee flexors					
(%)	10.6 (8.8)	18.1 (10.7)	0.887 (0.05)	<0.001 (1.0)*	0.3 (0.17)
M	9.5 (9.0)	18.8 (13.1)			
F	11.7 (8.7)	17.5 (8.1)			

* Significant main effect or interaction (p<0.05)

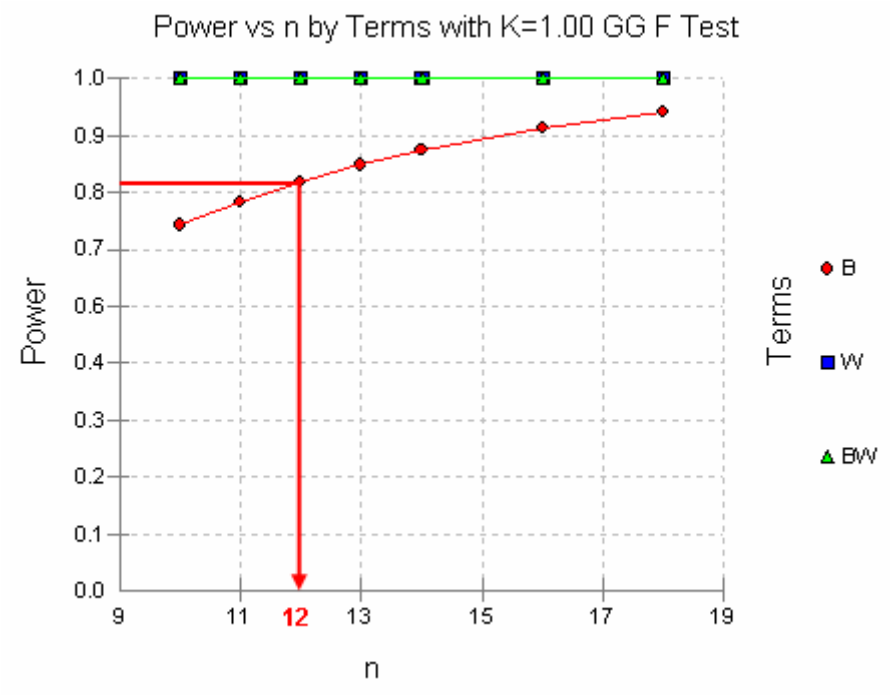

Figure 2-1. Correlations of power and sample size for each combination of variables. Note: B (between-subject variable), W (within-subject variable), BW (interaction between B and W).

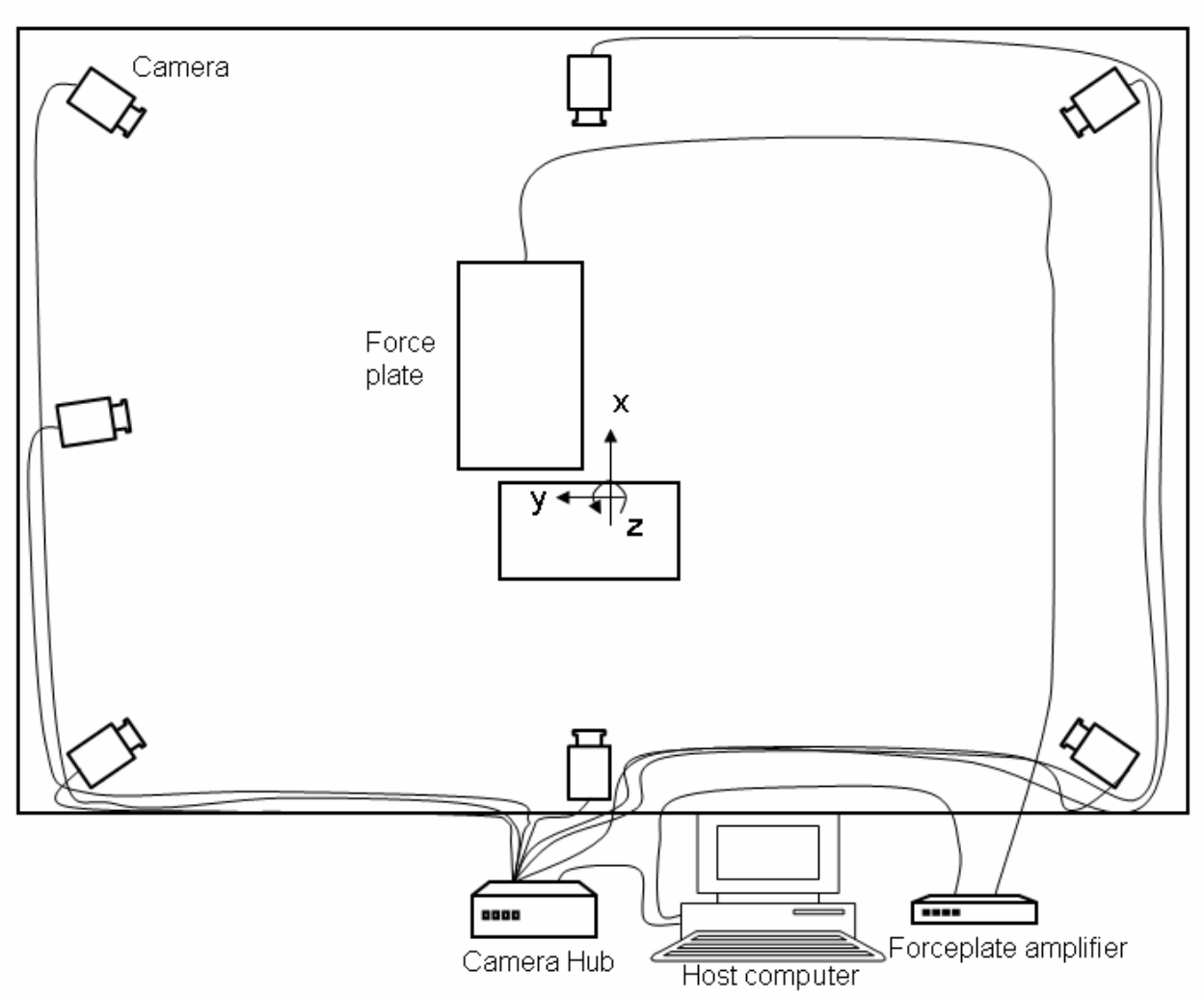

Figure 2-2. Experimental setup.

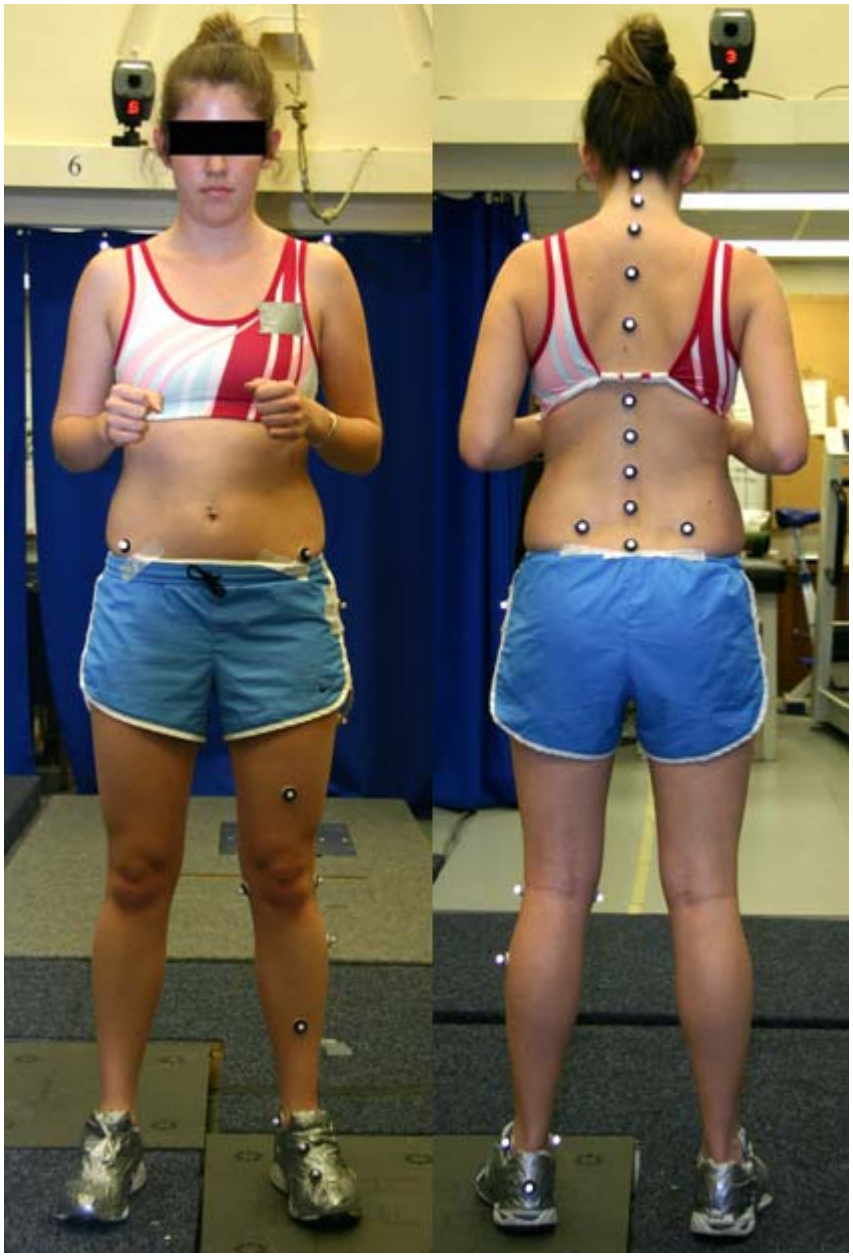

Figure 2-3. A subject with markers on.

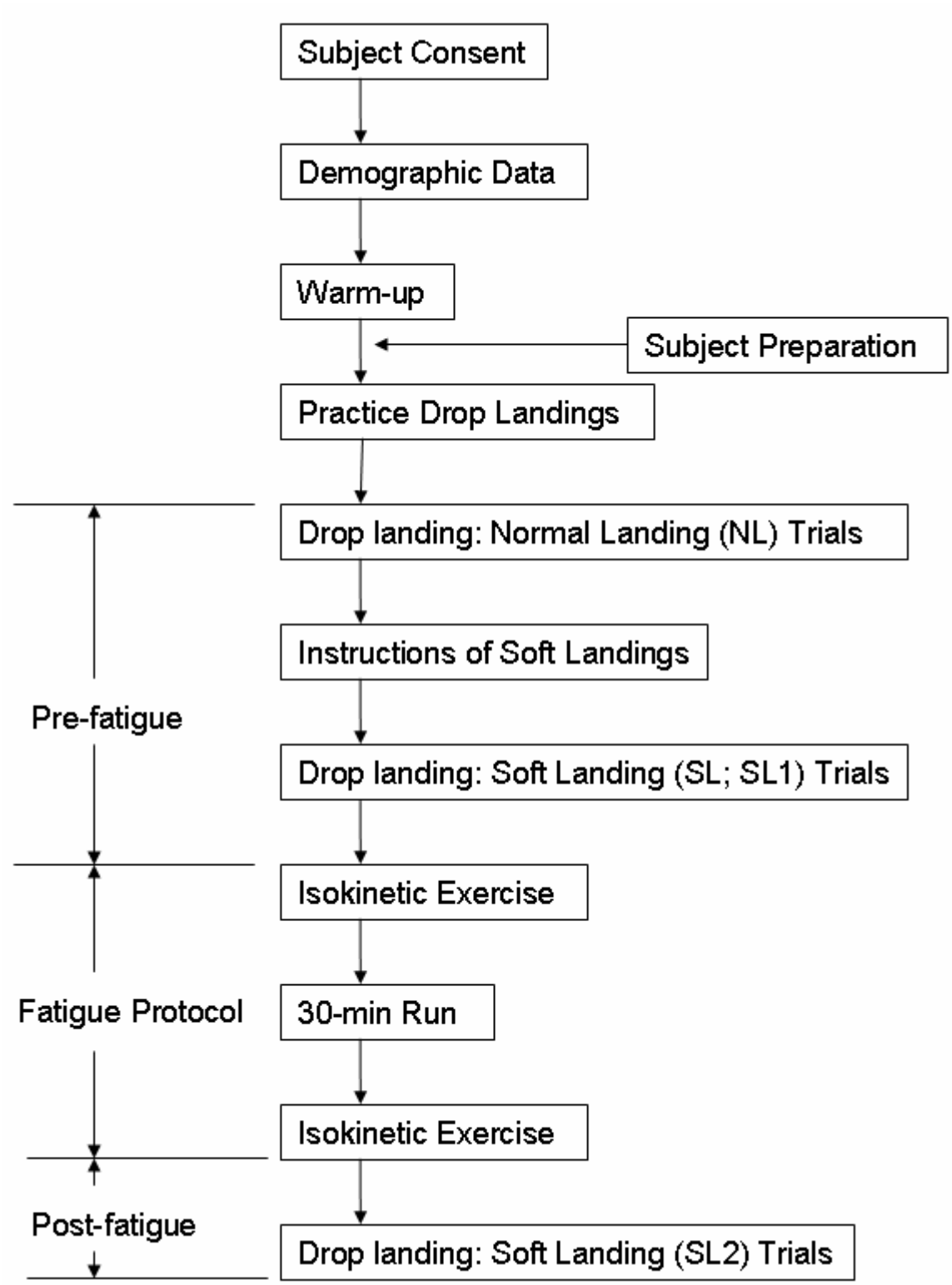

Figure 2-4. Overview of the experimental procedures.

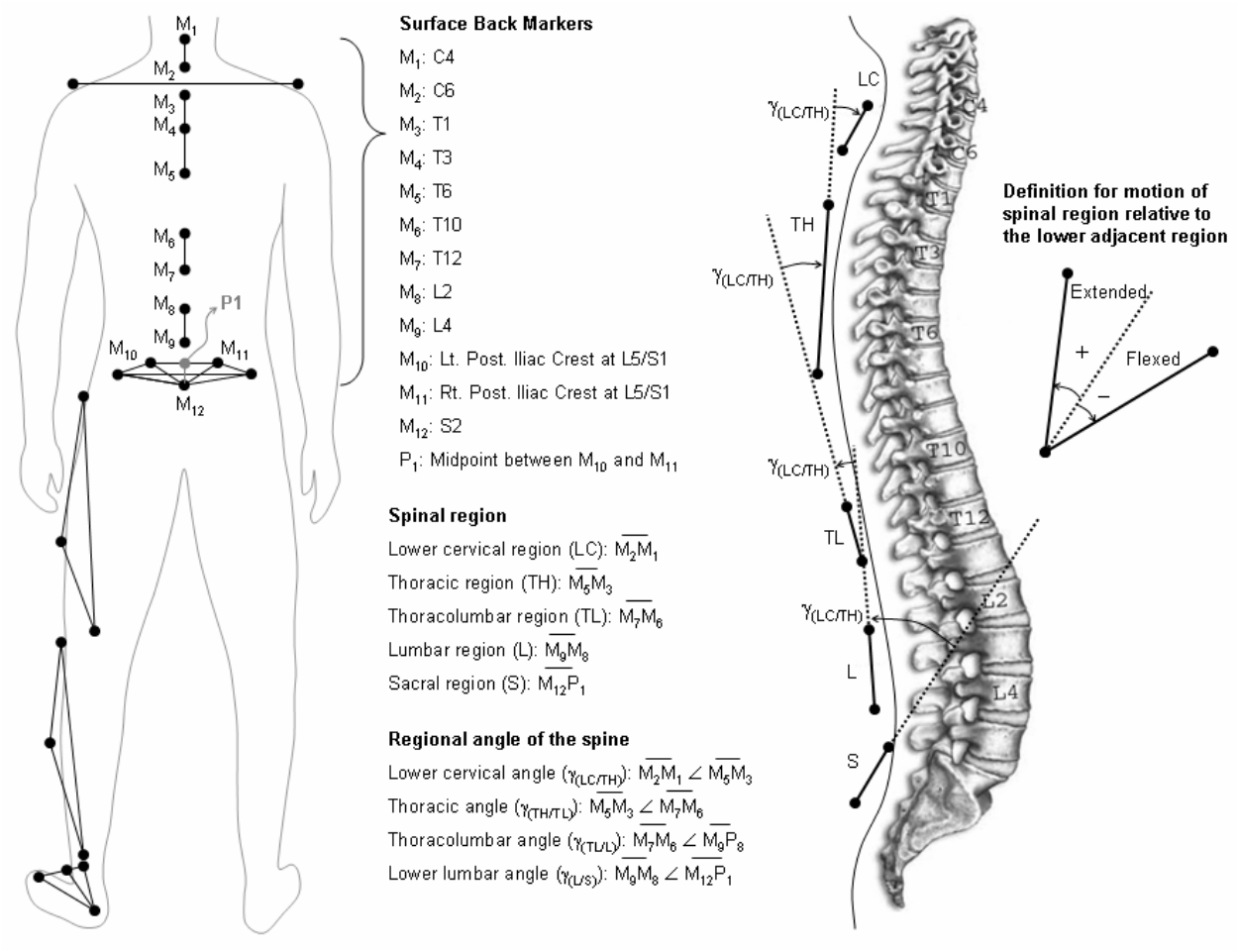

Figure 2-5. Marker placement (left) and definition of regional angles of the spine (right).

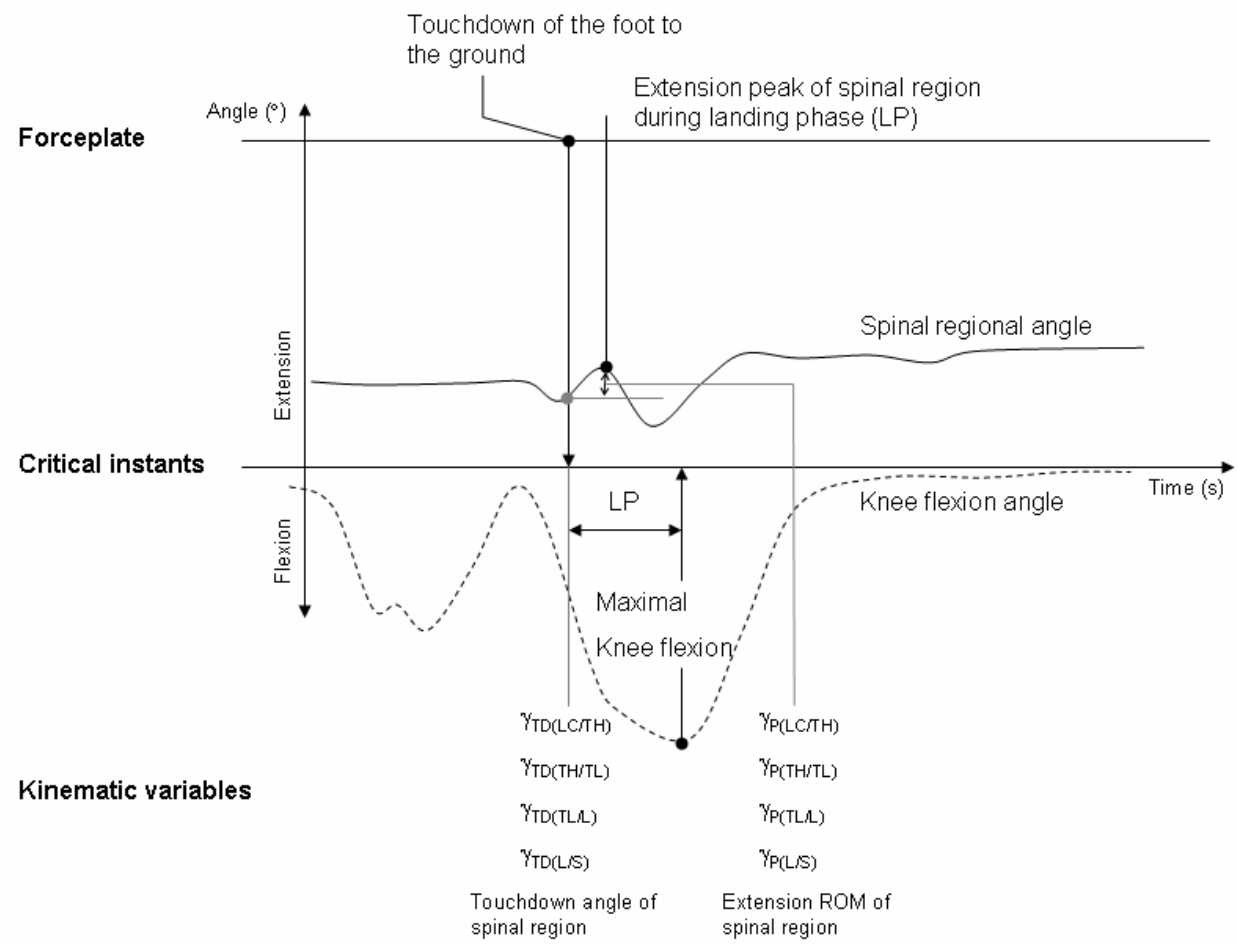

Figure 2-6. Kinematic variables defined by the critical instants identified from kinematic and forceplate data.

CHAPTER 3
LITERATURE REVIEW

The purpose of this study was to investigate the mechanical characteristics of the spine during drop landings using different landing techniques and fatigue status of lower extremity muscles. A biomechanical model of the lower extremity and spine was employed to study the kinetics and kinematics of the spine and lower extremity joints. Research investigating spinal mechanics using *in vitro* and *in vivo* techniques is quite extensive while studies of spinal mechanics in jumping and landing are quite limited. Previous *in vitro* research focused on range of motion of spinal segments in various conditions. Previous *in vivo* research focused on the spinal mechanics during common, everyday activities without considering lower extremity activities. More specific knowledge about spinal mechanics during vigorous physical activities is necessary to have a better understanding of spinal mechanics when the spine is under dynamic loadings. A description of spinal mechanics during drop landings is not currently available in the literature.

Early studies to investigate spine biomechanics were mostly focused on defining mechanical properties of spinal structures and segments in various conditions using different instrumentations. Recent *in vitro* research focuses on the cause-and-effect relationship and mathematical modeling of specific situations with cadaveric spinal segments using computational programming and statistical procedure to explain the complicated situation and to develop the spinal instrumentation by minimizing and simplifying the mechanical conditions. Recent *in vivo* research focuses on estimating and verifying the results from *in vitro* and clinical studies using optoelectrical systems, EMG, forceplate and so on.

To assist the reader in understanding the current issues of research on spinal mechanics, the literature review will be presented under the following headings: biomechanical properties of

spinal structures, biomechanical properties of spinal segments, biomechanical performance of spine *in vivo*, biomechanical etiology of spinal pain, biomechanical performance of painful spine, and landing biomechanics.

Biomechanical Properties of Spinal Structures

The mechanical properties of the bony structures (including facet joints), ligaments, spinal muscles and intervertebral disc of each spinal level have been thoroughly identified in previous studies. The intervertebral disc has received more attention than the other spinal structures due to its specific anatomical and biomechanical features. Defining disc characteristics is the first step in reviewing the mechanical properties of the spine.

Intervertebral Disc

An intervertebral disc consists of three components: the nucleus pulposus, the annulus fibrosus, and the cartilaginous end-plate. However, there is no clear landmark to differentiate the nucleus pulposus and annulus fibrosus because the peripheral region of the nucleus pulposus merge with the inner region of the annulus fibrosus. The nucleus pulposus is a centrally located mucoid material in semi-fluid state, and its water content of which ranges from 70-90% (Panagiotacopulos et al., 1987). The annulus fibrosus forms the outer region of the disc, and consists of collagen fibers in a highly ordered pattern. The collagen fibers are layered in 10-20 sheets called lamellae and arranged in a helicoid manner. They run in the same direction in a given lamella but in opposite directions in adjacent lamella. The lamellae are thick in anterior and lateral regions of the annulus, and thin posteriorly (Inoue, 1981). The end-plate is composed of hyaline cartilage about 0.6-1.0 mm thick which separates the other two components of the disc from the vertebral body (Roberts et al., 1989). The end-plate covers the entire nucleus pulposus, but does not cover the entire annulus fibrosus peripherally. Instead, the ring apophysis, which is part of a vertebral body, covers the peripheral region of the annulus fibrosus. Because of the

attachment of the annulus fibrosus to the end-plate, the end-plate is strongly bound to the disc, but weakly attached to the vertebral body (Inoue, 1981).

The basic functions of the disc are to transmit loads from one vertebral body to the next and to allow movement between vertebral bodies. All components of the disc are involved in weight-bearing. When an axial compressive load is applied to a nucleus, the nucleus tends to reduce the height and expand radially towards the annulus fibrosus. This radial expansion exerts a pressure on the annulus which tends to stretch its collagen lamellae outwards. However, the tensile properties of the collagen resist this stretch, and the lamellae oppose the outward pressure exerted by the nucleus. Application of a 400 N load to an intervertebral disc causes only 1 mm of vertical compression and only 0.5 mm of radial expansion of the disc (Hirsch & Nachemson, 1954). The nucleus pressure is also towards the end-plates, and constrained by the end-plates and vertebral bodies. The pressure on the end-plates serves to transmit the part of applied load from one vertebra to the next, and the radial pressure on the annulus fibrosus braces it and prevents the annulus from buckling (Roaf, 1960b).

Brown and colleagues (1957) conducted static tests to compare the relative strength of the disc with that of the vertebral body in compressive loads, without the posterior elements. They found the first structure to fail in such a construct was the vertebra, instead of the intervertebral disc, because of the fracture of the end-plates. They also observed no difference between the vertebrae with normal discs and those with degenerated discs. The mode of failure by the pure compressive loads seemed to be mostly dependent on the condition of the vertebral body (osteoporosis of the vertebrae), not on the condition of the disc.

During distraction, all points on one vertebral body move an equal distance perpendicularly from the upper surface of the other vertebral body. Consequently, every collagen

fiber in the annulus fibrosus is equally strained, and resists distraction. However, the disc is not often subjected to the tensile loads under normal physiologic activities. Also, even under the distraction of the spine, the discs are under the compression load due to spinal muscle activity. However, the annulus fibrosus is subjected to tensile stresses in various physiologic activities. In addition to compression, any direction of the bending (flexion, extension, and lateral bending) moves the instantaneous axis of rotation to lie outside of the disc and the disc is subjected to the tensile stress at the opposite side of the bending (White & Panjabi, 1990).

Bending involves lowering one end of the vertebral body and raising the opposite end. This causes distortion of the annulus fibrosus and the nucleus pulposus. In forward bending, the anterior annulus is compressed and the disc tends to bulge anteriorly. The nucleus pulposus is also compressed anteriorly, but the elevation of the posterior end of the vertebral body relieves the pressure on the nucleus posteriorly (Brown et al., 1957; Shah et al., 1978). However, Roaf (1960a) did not find any changes in shape or position of nucleus pulposus on the nucleographs of the disc during flexion/extension. This supports the relevance of maintaining a slightly flexed lumbar spine posture as a treatment and prophylaxis for the patients with low back pain. The increase in disc pressure observed *in vivo* during bending of the lumbar spine may not be just from bending, but from the result of compressive loads applied to the discs by the action of the spinal muscles which are involved in bending motion (Ortengren et al., 1981).

During torsional movement of the inter-body joint, only the collagen fibers in the annulus in the direction of movement have their points of attachment separated. Thus the annulus fibrosus resists torsional movements with only half number of lamellae. Farfan et al. (1970) conducted experiments using cadaveric vertebra-disc-vertebra construct including posterior structures to examine the effect of torsional load. They found that the failures occurred at the

annulus in the final phase of the loading. The average failure torque for the normal discs was found to be 25% higher than that for the degenerated discs, and the average torsional angles at failure were 16° and 14.5° for normal and degenerated discs, respectively. Torsion of inter-body joints seems be the most likely mechanical factor to injure the annulus.

In pure shear movements of the inter-body joint, only half of the fibers in the annulus are strained, and the shear stress is raised mostly at the side of the loading. The shear stiffness in the horizontal plane was found to be about 260 N/mm, and this could be the large force to cause an abnormal horizontal displacement in the normal disc (White & Panjabi, 1990). This means that it is rare for the annulus to fail clinically due to pure shear loading.

In addition to the load characteristics, the intervertebral disc demonstrates the time dependent behavior which is called the viscoelastic property: creep and hysteresis. If a constant force is applied to a viscoelastic structure for a prolonged time, further movement will be detectable after the end of physiologic motion. If this movement is small in amplitude, occurs slowly, almost imperceptible, then it is called as creep. Kazarian (1975) performed compression creep study on functional spinal units (FSUs) and differentiated the disc specimens into four grades according to the degree of degeneration. He found the creep and degeneration grade of the disc are related. The non-degenerated discs creep slowly and reach their final deformation after a long time, compared with the degenerated discs. This means that degenerated disc loses the capability to attenuate shock and to distribute the load uniformly over the entire end-plate.

Viscoelastic structures also show differences in mechanical behavior during loading and unloading. Restoration of the initial length of a structure from unloading occurs at a slower rate and to a lesser extent than did the deformation from loading. This difference in mechanical behavior is referred to as hysteresis, and reflects the amount of energy lost compared with

structure at initial loading. Virgin (1951) observed that hysteresis is largest in young people and smallest in the middle-aged ones, and the lower thoracic and upper lumbar discs show less hysteresis than the lower lumbar discs. He also found hysteresis decreased when the same disc loaded repeatedly. This means that the disc is less protected against repetitive loads.

When forces are repeatedly applied to a material, it does not behave the same way each time. Each application produces a certain amount of hysteresis, and the material is altered slightly. Following many repetitions, small weaknesses accumulate and weakness in the material becomes apparent. After several frequent repetitions of a stress, the material may fail at a certain stress which is less than that required to damage the material following a single application of a force. This is referred as the fatigue failure, and the fatigue tests of the disc were developed to identify the number of load cycles that can be tolerated before disc failure develop. Brown et al. (1957) conducted a fatigue test on the disc with a small constant axial load and a repetitive forward bending of 5°. The disc failure started to occur after 200 cycles of bending, and complete failure occurred after 1000 cycles. However, the fatigue tolerance of the disc *in vivo* is not known.

Vertebra

The basic morphology of the vertebrae in various regions of the spine from C3 to L5 is approximately the same. The size and mass of the vertebrae increase from C1 to L5 vertebra. This is the mechanical adaptation of the vertebrae to the progressively increasing compression loads. Chalmers et al. (1966) observed that the strengths of vertebral cancellous bone of each lumbar segment are almost the same. This means the variation of the vertebral strength according to the spinal level is mostly due to the size of vertebrae. However, vertebral strength decreases with age. Bell et al. (1967) found that a small loss of osseous tissue produces much loss of

vertebral strength: a 25% loss of the osseous tissue results in a more than 50% decrease in the vertebral strength. The bone mineral density of female vertebrae is less than that of male at any corresponding age, but the rate of decrease is not different between males and females (Hansson & Roos, 1980).

The vertebral body is designed for load-bearing of large compression. First, the vertebral body is not a solid bone block, but a shell of cortical bone and cancellous core. Second, the space between the trabeculae in the cancellous core can be used as the channels for the blood supply and venous drainage of the vertebral body. Likewise, the presence of bone marrow in the intertrabecular spaces acts as a useful element for transmitting the loads and absorbing the force (White & Panjabi, 1990). Rockoff et al. (1969) conducted compression strength tests on two groups of vertebrae without posterior elements: the vertebral bodies with central hollow and the other ones without outer shell. They found the loss of strength in both specimens, and the sharing of the compressive load by the cortical shell was about 65%.

Facet joints are important stabilizing structures and carry about 18% of total compressive load in the lumbar spine region (Nachemson, 1960). However, King et al. (1975) pointed out that the load-sharing between facet joints and disc is much complex in their dynamics study with whole cadavers. They observed that the load sharing carried by the facets could be 0-33% and the value depended on the spine posture. Facet joints also contribute to torsional strength of FSU. Farfan (1970) observed that the vertebral body-disc-body with longitudinal ligaments share the torsional strength equally with the two facets and capsular ligaments, about 45% each. The remaining 10% was carried by the interspinous ligaments.

Spinal Ligaments

The ligaments from C2 to the sacrum are similar, and seven ligaments are generally referred to as the spinal ligaments in this region: the anterior and posterior longitudinal

ligaments, the intertransverse ligament, the capsular ligament, the ligamentum flavum, and the interspinous and supraspinous ligaments. The anterior and posterior longitudinal ligaments attach to the anterior and posterior edges of vertebral bodies and discs from basiooccipital to the sacrum and coccyx. These ligaments become deformed by the relative separation between adjacent vertebrae and by the bulging of the disc. Tkaczuk (1968) observed that the anterior longitudinal ligament was twice stronger than the posterior longitudinal ligament. The intertransverse ligament is known to have no mechanical significance in the lumbar region because of its small cross-sectional size (Chazal et al., 1985). The capsular ligaments contribute to the flexion stability in the cervical spine region (Panjabi et al., 1975). The ligamentum flavum has a high percentage of elastin (80%) when compared to the other ligaments. This allows a large extension without permanent deformation (Yahia et al., 1990). Nachemson and Evans (1968) found that the ligamentum flavum has pre-tension, and this produces resting compression of the disc. Consequently, the high elasticity and pre-tension of the ligamentum flavum minimizes the chances of any impingement to the spinal canal during sudden spine motions.

Biomechanical Properties of Spinal Segments

Multisegmental Mechanics of the Spine

The basic motion segment of the spine is referred to as the functional spinal unit (FSU), which consists of two adjacent vertebrae, intervertebral disc, and the connecting ligaments without the spinal musculatures. Generally, a FSU exhibits similar biomechanics to those of the entire spine, and can be used as a common testing specimen *in vitro*. The range of motion (ROM) of a FSU is represented by the sum of two distinct phases: neutral zone and elastic zone. The neutral zone is defined as the low-load response of FSU near the neutral position, and the elastic zone is defined as the spinal behavior beyond the neutral zone up to the end of the physiologic limit (Figure 3-1) (White & Panjabi, 1990).

Generally, the neutral zone is referred to as a quantitative measurement of the laxity around the neutral position of a FSU. It is known to increase with degeneration, surgical injury, repetitive cyclic loads, and high-speed trauma. In flexion/extension and lateral bending, the neutral zone is the largest in the lower cervical region. In axial rotation, the neutral zone is the largest in the C1-C2 region (Table 3-1).

By defining the neutral and elastic zones in the load-displacement curve of an FSU, the coefficient of flexibility and stiffness can be calculated. The flexibility coefficient is defined as the ratio of the displacement produced to the load applied. The stiffness coefficient is defined as the ratio of the resistance offered to the displacement imposed. However, the load characteristics of the spine is quite complex (nonlinear, biphasic, and viscoelastic) and cannot be demonstrated by a single number. Previous studies showed that FSUs are more flexible in tension than in compression in all regions of the spine. The shear flexibility is not quite different in each direction (anterior, posterior, or lateral). The spine is more flexible in flexion than in extension in all regions except the sacroiliac joint. Flexibility values for lateral bending are in between the values of flexion and extension (Panjabi et al., 1988; Panjabi et al., 1976).

Axial rotation is generally known to be more harmful to the disc than the other motions, except for a combination of axial rotation and lateral bending (Farfan et al., 1970). In the cervical region, the spine is about 37% as flexible in torsion as compared with flexion. In the upper thoracic region, the torsional stiffness is about the same as in flexion. The torsional flexibility of the lumbar region is about 27% of flexion flexibility, which is the lowest value in all regions. However, the torsional flexibility is much greater at the lumbosacral joint (55% of flexion) and sacroiliac joint (250% of flexion) (White & Panjabi, 1990).

Individual physiologic motions of FSUs are inherently connected, and which is called the coupling. Coupling of spinal motions is due to the geometry of individual vertebrae, connecting ligaments, discs, and the curvature of the spine. The motion produced by an external load is defined as the main motion, and the accompanying motions are called the coupled motion. In the thoracic region, there is a strong coupling between all the motions in the sagittal plane (translation and rotation) (Panjabi et al., 1976). The coupling of axial rotation with lateral bending is a very common physiologic motion in the cervical and lumbar regions (Moroney et al., 1988; Panjabi et al., 1977).

Regional Mechanics of the Spine

The entire spine is divided into cervical (upper C0-C1-C2, middle C2 – C5, lower C5 – T1), thoracic (upper T1 – T4, middle T4 – T8, lower T8 – L1), lumbar (L1 – L5), lumbosacral (L5 – S1) and sacroiliac regions, based on the kinematic, kinetic, and clinical characteristics.

The upper cervical region is composed of occipital-atlanto-axial joints (C0-C1-C2) and is the most complex region of the spine, anatomically and kinematically. Most of the axial rotation and some of the flexion-extension and lateral bending of the head come from the upper cervical movements. The dominant atlantooccipital (C0-C1) motion is mostly flexion/extension, some lateral bending, and tiny axial rotation (Table 3-2). The atlantoaxial (C1-C2) articulation consists of four joints: two atlantoaxial lateral joints, the atlantoaxial median joint (between anterior arch of the atlas and dens axis), and a joint between the posterior surface of the dens and the transverse ligament. The lateral atlantoaxial joint capsule is loose and allows a great deal of axial rotation, in which the vertical axis of dens acts as a pivot about the atlas rotation. The possible atlantoaxial motions are also summarized in Table 3-2.

The anatomical structures and the function of the middle and lower cervical regions are quite different to those of the upper cervical region. The dominant motion in the lower cervical

spine is flexion/extension. Lysell (1969) observed the routes of each cervical vertebra in the sagittal plane from flexion to extension or vice versa. The movement is a combination of translation and rotation, and he called that movement the 'top angle', which indicated the arch steepness of the route, generated during the flexion/extension. The arches were flat at C2, steepest at C6 and followed by C7. The average ROM in healthy adults in the middle and lower cervical regions are summarized in Table 3-3.

Dvorak et al. (1992) performed an *in vivo* test to measure the cervical ROM, based on the modified inclinometer technique, to define the age- and gender- related differences. They found a general tendency of decreasing cervical ROM as the age increased. The most drastic decrease in cervical motion occurred at the age of 30-39 and 40-49 years. The cervical motions that did not decrease with age were the rotation out of maximum flexion and the upper cervical rotation. Females showed greater ROMs in all planes of cervical motion. However, there were no significant differences between genders for the group over 60 years. The typical cervical ROM values are presented in Table 3-4.

The human thoracic spine is a unique spinal region to be adapted to an erect posture and load-bearing. The predominant posture of the thoracic spine is a kyphotic curve while the last region (T11 – L2) is almost straight in the sagittal plane. Thoracic kyphosis may arise from postural and structural factors. Postural factor come from positioning of the spine due to the ligamentous tension and muscle tone, as well as the disc configuration. The shorter ventral height of thoracic vertebral bodies than dorsal one contributes to the structural kyphotic curve in the thoracic region. Due to the kyphotic curve of the thoracic spine, the axial load applied to this region generates a bending moment to cause further flexion. As a stability of this region, dorsal tension-band capacity by the posterior ligaments and ventral weight-bearing by the vertebral

bodies are the vital combination to prevent spinal deformities in this region (Benzel & Stillerman, 1999).

There are two transitional regions in the thoracic spine: cervicothoracic and thoracolumbar junctions. The biomechanical characteristics of these junctional regions are described as a blending of two adjacent regions. The upper thoracic region has a very limited flexion/extension, whereas the caudal region from T10 has a larger range of flexion/extension. The sagittal orientation of the facets in the lower thoracic region severely limits axial rotation and, to a lesser extent, laterals bending. The facets of the upper thoracic spine are similar in orientation to those of the cervical spine, and similar motion characteristics occurred at the cervical and upper thoracic regions. Likewise, the facets of the lower thoracic spine are similar to those of the lumbar spine and similar motions are seen in both lower thoracic and lumbar regions. Representative ROMs of different motion segments of the thoracic spine are summarized in Table 3-5.

The pattern of motion in the sagittal plane for the thoracic spine is similar to that in the cervical spine. To describe the motion of the thoracic vertebra in the frontal and sagittal planes, the 'top angle' was also employed. In the sagittal plane motion, the arch is quite small, and there is no variation according to the level. The arch in the frontal plane is also flat, but greater than that in the sagittal plane. Also in the frontal plane, the arch tends to increase from T1 to T12. There is also coupling of lateral bending and axial rotation in the thoracic region. The pattern of coupling in the thoracic region is similar to the one in the cervical region. However, the coupling pattern in the lower thoracic region is not as strong as that in the cervical region (White, 1969).

Movements in the thoracic spine are greatly limited by the facet orientation and by the rib cage. The costotransverse and costovertebral articulations provide strong, stable attachment of

the thoracic vertebrae to the ribs. The costosternal articulations also contribute to the stability of the thoracic spine (Pal & Routal, 1987). Oda et al. (1996) found a significant increase of flexion/extension in thoracic motions after resection of the posterior elements. With the removal of the costovertebral joints bilaterally, large increases in lateral bending and axial rotation were observed. They concluded that the integrity of the costovertebral joints and the rib cage significantly contribute to the spinal stability of the thoracic region.

In the sagittal motion of the lumbar spine, there is a cephalocaudal increase in flexion/extension. The lumbosacral joint provide more sagittal plane motion than the other lumbar motion segments do. For the coronal motion, the ROM for each lumbar level is about the same, except for the lumbosacral region, which demonstrates limited lateral bending. Limitation of lumbosacral motion is about the same for the axial rotation. Representative ROMs of the lumbar spine are summarized in Table 3-6. Another important kinematic component in the lumbar spine is the sagittal plane translation. In general, 2.0 to 2.8 mm is referred to as the normal limit of anterior translation ofr a lumbar spinal vertebra, and 4.5 mm is an evidence of clinical instability for a lumbar motion segment.

There are several patterns of coupling motion in the lumbar spine. Pearcy et al. (1984) observed coupling of slight axial rotation and lateral bending with flexion/extension, based on their stereoradiographic study. Another coupling pattern is that of lateral bending and axial rotation. The direction of lateral bending with axial rotation in the lumbar region is that the spinous processes point to the same direction as the lateral bending. It is opposite to the pattern occurred in the cervical and thoracic regions. However, the coupling pattern of lateral bending with axial rotation at the lumbosacral joint is the opposite of that in the lumbar region and similar to the cervical and thoracic regions (Pearcy & Tibrewal, 1984).

The sacroiliac joint is partly synovial and partly syndesmotic. It is known to be completely ankylosed in 76% of the subjects over 50 years of age (White & Panjabi, 1990). However, studies of the motion about the sacroiliac joint have produced a wide range of results. Miller et al. (1987) performed a kinematic study of the sacroiliac joints in cadaveric specimens. They measured the displacements of the sacrum in relation to the ilium with each plane of loading. They observed that lateral bending of one side was 1.4°, anterior translation 2.74 mm, and axial rotation of one side 6.21° in their study. Walheim et al. (1984) observed 2-3 mm of vertical translation and 3° of rotation of the pubis at the symphysis pubis with one leg standing. Sturesson et al. (1989) also observed tiny motions of sacroiliac joints in their stereoradiographic study.

Biomechanical Performance of Spine In Vivo

Trunk Posturing

Trunk postures during various activities are related to the risk of developing a low back disorder (Granata & Wilson, 2001). Spinal compression below 3,400 N may be considered as a safe margin to prevent low back disorder in occupational population (Konz, 1982). However, a spinal injury associated with the instability can occur at a low compressive load (Granata & Marras, 1999). This means that the appropriate recruitment of the spinal and trunk muscles provide the stable support for the large load applied to the spine, but some postures may limit the ability of the muscles to maintain spinal stability (Wilke et al., 1995).

Nachemson (1966; 1981) conducted studies to verify the compression load applied to the disc *in vivo* by measuring the intradiscal pressure at L3/L4 level in various postures. The lowest compression was observed in the supine position (300 N), which is about 50% of that in standing (700 N) without external loads. During sitting without a back support, the compression went up

to 1,000 N. Forward bending of the trunk generated the largest compression force to the spine up to 1200 N. Ledet et al. (2005) measured intradiscal loads in baboons *in vivo* and found similar results to Nachemson's. Takahashi et al. (2006) also performed the same testing at L4/L5 level with young, healthy subjects, and found similar results. However, their values in specific postures are different from those reported by others. The values of compression loads in different studies are summarized in Table 3-7.

The data from Takahashi et al. (2006) were greater than those from Nachemson (1966; 1981). Takahashi et al. explained the differences with several reasons. Their subjects were all young and normal subjects. Their techniques seemed to be much more developed than the old one used in Nachemson studies. The specific level used in intradiscal measurement was lower than that in Nachemson studies. Takahashi et al. concluded that the risk of intervertebral disc injuries or degeneration could be induced by a simple repetitive forward bending of the trunk in everyday movements. However, slightly flexed, relatively straight or nonlordotic position of the lumbar spine during standing is used more frequently in the populations who complain less often about back pain. Also, the flexion of the hip reduces the tension of the psoas muscles and the lordosis of the lumbar spine, resulting in reduced load on the lumbar spine.

In addition to the forward bending of the trunk, twisting, lateral bending and asymmetric posture combined with lifting are also known to increase the risk of low back problems (Kelsey et al., 1984; Marras et al., 1993). Some EMG studies explained that the high loads to the spine in the unstable and asymmetric postures are from the co-contraction of agonistic and antagonistic spinal muscles (Cholewicki et al., 1997).

Weight Lifting

Lifting and bending episodes accounted for 33% of all work-related causes of back pain (Damkot et al., 1984). Increasing weights anterior to the spinal column greatly increases the

forces which are exerted on the lumbar spine. This is due to the forces developed in the spinal muscles in order to maintain the equilibrium. The resultant forces applied to the fulcrum, which is the lower lumbar region, are very high.

The distance of the weight from the body is directly related to the joint reaction force (high disc pressure) to the lower back, the greater force required by the erector spinae muscles (high electromyographic activity), and a need of greater truncal support to protect the spine (high truncal pressure). This indicates the significance of closer distance of the object to the body as the proper lifting technique. Another factor to determine a proper lifting is the back posture. For the optimal lifting methods, the squat lift (knee bent and back straight) is generally considered to be safer than the stoop lift (knee straight and back bent) in bringing the load closer to the body, and reducing the back muscle demands to counterbalance the additional moments. These techniques consider the posture of the back in addition to the distance of the objects. However, many workers prefer the stoop lift over the squat lift. There is an increased physiologic cost and more rapid fatigue development in a squat lift. And the squat lift is not always possible because of the lift setup and load size. Likewise, the risk of developing low back pain by lifting tasks depends rather on the lumbar posture than the choice of lifting techniques (van Dieen et al., 1999).

There is a conflict about the favorable lumbar postures during lifting tasks. Some advocate lordotic and straightening lumbar posture because they believe increased erector spinae activity is beneficial in augmenting spinal stability and decreasing anterior shear force on the spine (Hart et al., 1987). However, others favored the kyphotic lift (flexed lumbar spine), because they believe passive ligaments of the lumbar spine can relieve the active extensor muscles (Gracovetsky et al., 1981). Cholewicki et al. (1992) tested professional power lifters to

evaluate the kinematics of the lumbar spine and resultant posterior ligament lengths during lifting tasks. They observed significantly smaller lumbar flexion and increased lengths of the ligaments during the lifting when compared with the full flexion of the trunk. They concluded that the back muscles were substantially responsible for resisting trunk flexion moments during heavy lifting.

Arjmand and Sirazi-Adl (2005) tested the kinematics of the lumbar spine and activity of selected spinal and trunk muscles in healthy subjects during a static lifting. They examined the lumbar spine postures (lordosis, kyphosis) during the lifting procedures and found the lordotic posture increased extensor muscle forces, axial compression and shear forces at L5/Sl. They recommended the moderate flexion posture of the lumbar spine as a posture of choice in static lifting tasks. Lifting capacity is generally used to determine the degree of spinal impairment state, and the back strength and aerobic capacity are known to be the contributing factors (Matheson et al., 2002).

Sitting and Standing

The back rest and lumbar support is known to decrease the loads applied to the spine during sitting. Andersson et al. (1977) performed a study to estimate intradiscal pressure of L3/L4 under different backrest inclinations and lumbar support conditions. They found the highest intradiscal pressure in the sitting with no lumbar support and a 90° backrest inclination, and the lowest disc pressure and the least electromyographic activity of the paraspinal muscles in the sitting of 120°-inclination and 5-cm width of lumbar support. More specifically, the lumbar support has the greater influence on lumbar lordosis and the backrest inclination has more influence in reducing the loads on the lumbar disc (Andersson et al., 1979). The arm rest is also known to reduce intradiscal pressure (Kelsey & Hardy, 1975).

The ability to stand up from sitting is a prerequisite for walking and other independent function. Sit to stand demands coordinated movements of linked body segments to transport the center of body mass in a horizontal then vertical direction while maintaining balance over a small base of support, the feet. Previous studies reported these kinematic characteristics of standing from sitting: flexion of the trunk and hips bring the center of mass forward, followed by bilateral extension of the lower limb joints, and trunk extension to raise the body in a vertical direction over the feet (Doorenbosch et al., 1994). Tully et al. (2005) studied the kinematics of the body segments including the thoraco-lumbar region during standing from sitting. They divided the standing movement into phases before and after the lift-off, based on the relation of position between the buttock and the sitting object. Before the lift-off, they observed a forward leaning of the trunk, accomplished by concurrent lumbar and hip flexion (1:3). As the lumbar spine flexed the thoracic spine extended, resulting in a trunk angle of 45.7° at lift-off with respect to the horizontal plane. Following the lift-off, the hip and lumbar spine extended and the thoracic spine flexed, with the standing thoracic angle approximating the initial thoracic posture in sitting.

Walking

The biomechanical function of the trunk during walking has been investigated extensively. Earlier studies examining the trunk kinematics during walking considered only the entire trunk motions with respect to the pelvis. Later studies examined the movements of the lumbar and thoracic spines or pelvis. Crosbie et al. (1997) studied the patterns of spinal motion during walking using a model including upper and lower trunk, lumbar and pelvis segments. They used three spatial surface markers in each spinal segment on the back surface of the subjects. The pattern of flexion/extension of each segment was generally biphasic throughout the gait cycle. The pelvis rotated into negative pelvic tilt at heel strike. This was followed by a counter-motion

to a maximum positive pelvic tilt in the single support phase. The lumbar spine reached maximum flexion at heel strike. This was followed by a rapid extension to neutral until the single support phase. The lower thoracic segment extended maximally at heel strike, and returned to a neutral at mid-stance, then extended again through the late stance. The counter-motions occurred between the lumbar and lower thoracic segments at heel strike. They concluded that spinal segments demonstrate complementary movements to the motion of the pelvis, and pelvic motion responds to the need of advancing the lower limb and transferring the body weight from one supporting side to the other. Lumbar ROMs during walking and running are summarized in Table 3-8.

Thousands of repetitive low level loadings are applied to the spine in everyday activities. During normal walking, activation of the spinal muscles, acceleration of the trunk, combined with the external loadings result in cyclic spinal loads. Some studies investigated the magnitude of these loads, in conjunction with the spinal motion and muscular activities during walking. Callaghan et al. (1999) conducted a biomechanical study using two models to estimate loads applied to the L4/L5 level: linked segment model with EMG technique and rigid segment model with inverse dynamic technique. The joint loading (at L4/L5) calculated by the EMG model resulted in large increases in the maximum compressive forces, compared with the joint reaction forces calculated using inverse dynamics. Including the muscular component resulted in a more than three-fold increase in joint load. The compression loads applied to the lower lumbar level during walking from different studies are summarized in Table 3-9.

Unlike the differences in compressive load, the joint shear forces (anterior/posterior, lateral) obtained using the two techniques were similar. The flexion/extension moment had two peaks throughout the gait cycle. At heel contact there was a peak flexor moment followed by a

peak extensor moment around toe-off. During a faster speed gait, the flexion/extension moment at L4/L5 shifted to the extension side and demonstrated a high extensor moment around toe-off. The lumbar spine motion with respect to the pelvis was quite consistent within and between subjects. The sagittal motion of the lumbar spine showed several dominant phases. Following heel contact a flexion phase was present until the relative spine motion reached maximum flexion just following toe off. And the spine remained in a constant posture and additional extension phase during the single stance. They concluded that the loads and motions for the lumbar spine during gait depended on the walking speed. Increasing walking speed increases the lumbar spine ROM, activation of spinal and trunk muscles, and anterior/posterior shear forces.

Running

Running typically requires the spine moving through only a limited range of motions. Acute injuries to the spine directly from running activities are relatively infrequent and amounted approximately 11 – 13% of all injuries sustained (Walter et al., 1989). The frequent and significant spine injuries related to running are largely due to the repetitive axial compressive loading of the spine which occurs during the foot strike in each stride. A typical distance runner who runs 130 km per week in training might subject to about 40,000 foot strikes per week (Cavanagh & Lafortune, 1980). Several case studies have highlighted the overuse injuries of the lumbar spine and pelvis in running (Guten, 1981; Koch & Jackson, 1981).

Schache et al. (2002) conducted a kinematic study of lumbar spine and pelvis during running. The average ROMs of the lumbar spine and pelvis on each plane are summarized in Table 3-8. They found significant inverse correlations between flexion/extension of the lumbar spine and anterior/posterior tilt of the pelvis and lateral bend of the lumbar spine with obliquity of the pelvis. Essentially, as anterior tilt of the pelvis increased during the terminal stance, extension of the lumbar spine also increased. When the lumbar spine was laterally bent to the

same side of the foot contacted to the ground, the pelvis was lowered on the opposite side. When the lumbar spine began to laterally bend towards the opposite side of the foot contacted, the pelvis began to elevate on the opposite side of the foot contacted. They also found a significant positive correlation between the axial rotations of lumbar spine and pelvis. However, coordination between the axial rotations of the lumbar spine and pelvis was out of phase by 21% of the running cycle.

The kinematic pattern of axial rotation of the pelvis during running is different to that during walking. At the initial contact of one foot during walking, the pelvis showed maximal rotation to the opposite side (Whittle & Levine, 1999). This movement helps in augmenting the stride length at that time. With the loss of the double support phase during running, the pelvis along with lower extremities are no longer required to be engaged in a stride lengthening mechanism. At the initial contact during running, the pelvis was rotated to the same side of the foot contacted. This movement was suggested as minimizing the horizontal braking force at the initial contact to avoid potential loss of running speed (Novacheck, 1998; Schache et al., 1999).

Schache et al. (2003) performed another kinematic study of lumbo-pelvic-hip complex during running to define the gender differences. They found that females displayed a shorter stance time, swing time, stride time and stride length, and a higher stride rate than males. Mean waveforms were different in the peak-to-peak oscillations and the offset of pelvis anterior/posterior tilt. Females displayed greater amplitudes of lumbar spine lateral bend and axial rotation, pelvis anterior/posterior tilt, obliquity and axial rotation, and hip adduction/abduction than their male counterparts. The mean positions of anterior pelvic tilt across the running cycle were 20.2° for females and 16.9° for males. The prevalence of pelvic-

femoral stress fractures in female runners might be explained by these findings (Bennell et al., 1996; Pavlov et al., 1982).

Wilke et al. (1999) studied the intradiscal pressure of non-degenerated disc at L4/L5 in a 45-years old man during various activities. They observed intradiscal pressure of 0.5 MPa during relaxed standing and 0.35-0.85 MPa during jogging with tennis shoes. Rohlmann et al. (2001) also recorded a peak intradiscal pressure 0.85 MPa while jogging on the treadmill of, which was 170% of the pressure noted in standing. The intradiscal pressures recorded in different studies are summarized in Table 3-10.

There are several factors suggested in previous studies that affect the spinal posture and loading during running. Exhaustive running, as fatigue occurs, has shown biomechanical changes in the legs that lead to a lower effective body mass during heel strike (Derrick et al., 2002). This resulted in increased peak leg impacts and increased shock attenuation, which may change the spine loading forces with fatigue. Therefore, the loads applied to the spine may vary during a run, depending on the lower extremity activity and the level of fatigue (Lennard & Crabtree, 2005).

Another factor that affects the spinal loading during running is the type of shoes and insole materials utilized. An impulsive shock wave is generated at heel strike that is transmitted from the lower extremities through the spine. The use of shock absorbing insoles has been used to treat low back pain patients to lower the shock wave at low back level. On the other hand, the development of external force and the transmissibility of impact forces through the human body are increased by wearing soft soles. Ogon et al. (2001) performed a study to investigate the behavior of low back muscles during jogging barefoot and wearing identical athletic shoes. They observed that wearing shoes and insert materials decreased the rate of shock transmission to the

lower back and reduced the time interval between peak acceleration of lower back and peak spinal muscle response in jogging. It is from the increased latency between heel strike and peak acceleration at the lower back by wearing shoes. They suggested wearing shoes decreases the time interval between maximum external (peak acceleration at lower back) and maximum internal force (generated by the spinal muscles) in the lower back during running. Wen and associates (1997) found a significant correlation between leg length discrepancy and onset of lumbar pain within 12 months of running in a marathon training program. However, the clinical importance of leg length discrepancy in short distance running is not clear.

Biomechanical Etiology of Spinal Pain

Spinal pain can be caused by trauma, infection, tumor, and systemic diseases. However, the term 'spinal pain' is generally used to refer to the cervical, thoracic, and lumbar pain that is not related to these injuries and diseases. The common neck pains with or without arm pain and the back pain with or without leg pain which are frequently encountered in daily livings and cause the spinal degeneration are called spinal pain. Although there are specific considerations associated with spinal pain in different regions of the spine, there is much similarity in different spinal regions. Spinal pain usually occurs in the more mobile and lordotic portions of the spine and onsets at 30-50 years of age. Spinal pain occurs most often in the lumbar region, followed by the cervical region, with the lowest incidence in the thoracic region. Spinal pain can come from direct irritation of the nerves that innervate most of spinal structures. The posterior annulus fibrosus, the posterior vertebral body and the posterior longitudinal ligament are innervated by the sinu-vertebral nerve, which is considered the most common origin of spinal pain (Bogduk & Twomey, 1997). Another type of spinal pain is the indirect referred pain which is not fully understood and explained at this moment. Spinal pain is a major socioeconomic problem and approximately 80% of all back problems are of unknown origin (Vogt et al., 2001).

Specific motion, force, and high-quantity repetitive loading, or any combination of these may serve as mechanical stimulus to the spine, and which can be referred to as the abnormal mechanical causes of spinal pain and degeneration either quantitatively or qualitatively. There are many biomechanical factors known to contribute to spinal pain: vibration, lordosis, torsion, driving motor vehicles, material handling, leg length discrepancy, etc. However, there are still controversies on the roles of mechanical factors relative to spinal pain.

Vibration

Epidemiologic studies reported increased spinal pain and/or disc disease in those who drive more than 3 hours per day or operate vibrating equipment (Frymoyer et al., 1983). Vibration, particularly in the frequency domain of 5 to 15 Hz in which resonance of the spine can occur (Goel et al., 1994), is considered a key etiologic factor in low back pain (Magnusson et al., 1996), neurovestibular disorders (Seidel et al., 1988), and a causal factor in circulatory disorders such as Raynaud's syndrome (Dandanell & Engstrom, 1986). Thus, the industries such as the transportation and construction, as well as the military are working toward minimizing occupational exposure to potentially noxious mechanical stimuli (Bongers et al., 1988; de Oliveira et al., 2001).

Brumagne et al. (1999) performed a study to test the proprioceptive changes in response to vibration in human. They applied a vibration (70 Hz, 0.5 mm amplitude) to the multifidus muscle for 5 seconds and measured the lumbosacral repositioning accuracy. They found a significant increase in directional error during vibration of the paraspinal muscles. The subjects had the illusion during vibration that their pelvis was more posteriorly tilted, and accordingly, they undershot the target position. They explained their results with the reflex inhibition of the muscle spindle, which play an important role in proprioception. They concluded that vibration on the spine and trunk muscle can result in damage to or dysfunction of the muscle spindles.

Consequently, a decreased muscle spindle input could jeopardize spinal proprioception and segmental stability, and likely to make the spine more vulnerable to injuries and low back pain.

On the other hand, whole-body vibration has been utilized as an exercise therapy for musculoskeletal problems in sports, geriatrics, and rehabilitation laboratories (Bosco et al., 1999; Rittweger et al., 2000). Vibration is thought to elicit muscle activity via stretch reflexes (Clark et al., 1981). Rittweger et al. (2002) observed that metabolic power increased during the whole-body vibration from a ground platform (amplitude of 2.4 mm, 5.0 mm, 7.5 mm, frequency of 18 Hz, 26 Hz, 34 Hz, duration of 4 minutes) and that this metabolic power is augmented by the application of additional axial loads. Ritweger et al. (2002) performed another study on patients of chronic lower back pain devoid of specific spine diseases to verify the therapeutic effects of vibration exercise (maximum amplitude of 6 mm, frequency of 18 Hz, duration of 4 to 7 minutes). They compared whole-body vibration exercise with lumbar extension exercise, and observed a significant reduction in pain sensation and pain-related disability in both groups. Lumbar extension torque increased in the vibration exercise group, but significantly more in the lumbar extension exercise group. They concluded that a well-controlled vibration can be the cure rather than the cause of lower back pain.

Some animal studies indicated that brief (<20 min) daily durations of extremely low-level (0.5g), high-frequency (15-90 Hz) vibration can be strongly anabolic to the trabecular bone, increasing bone mineral density, trabecular width and number in the weight-bearing skeleton (Rubin et al., 2001; Rubin et al., 2002). These studies suggested the osteogenic potential of extremely low-level mechanical stimuli as a treatment for osteoporosis. Rubin et al. (2003) studied transmissibility of high-frequency (15-35 Hz) ground-based, whole-body vibration to the proximal femur and lumbar vertebrae. They observed 30-130% of transmissibility of the loading

vibration, regardless of the target region, frequencies, and posture of the subjects. In addition, transmissibility at the hip was different to the spine, mostly at the lowest frequencies in this study. For the loading frequency less than 20 Hz, the resonance at the hip exceeded 100% in the erect standing and relaxed standing postures. However, the resonance at the lumbar vertebrae was lower than that at the hip, and it was constantly maintained at the high frequency of loading vibrations and the other standing postures (relaxed standing, bend knee). They mentioned that slight changes in posture can have significant influence on the degree of vibration to be delivered to the spine or hip regions. They also emphasized the possible undesirable side effects of using whole-body vibration, against the prevention strategy for the osteoporosis. They suggested that vibration which approaches 1 g (9.8 ms^{-2}), even at beneficial high frequencies, should be avoided considering the risks to many physiologic systems.

Lordosis

Reduced or flattened lumbar lordosis has signified lumbar spine problems in previous studies (Adams & Hutton, 1985; Evcik & Yucel, 2003; Farfan et al., 1972). On the other hand, the cultural groups who spend considerable time in lumbar flexed position are known to suffer less low back pain and disc degeneration (Fahrni & Trueman, 1965). Although there are controversies raised in previous studies, the biomechanics of lumbar lordosis and back pain were found to be closely related in some instances. Frymoyer et al. (1984) performed a radiologic study to investigate the lumbar lordosis in low back pain patients. They found no association of lumbar lordosis with low back pain. Murrie et al. (2003) conducted MRI assessments of lumbar lordosis in between patients of low back pain and normal controls, and did not find any significant difference in both groups. Instead, they observed that lumbar lordosis is more prominent in women and those with a higher body mass index.

However, there are some cogent observations on the significant changes in the facet joints and discs associated with spine extension and hyperlordosis. Dunlop et al. (1984) reported that each degree of increased extension of the spine leads to a 4% increase in peak articular pressure in the facet joints. Yang and King (1984) reported that arthritic facet joints may bear up to 47% of the load transmitted. Thus, these increased loads to the facet joints lead to abnormal motion to the inferior facet articulation and damage to the joint structures, which can cause low back pain. The available evidence does not support strong conclusions, but there seem to be disadvantages to hyperextension of the lumbar spine if there is facet joint arthritis or disc pathology.

Roussouly et al. (2005) performed a radiologic study to classify the normal variation of sagittal lumbar spine and pelvis. They found that sagittal alignment of the human spine and pelvis was highly variable in different individuals in a standing position. The angle of the superior end plate of S1 with respect to the horizontal axis varied between 20° and 65°, the angle of global lumbar lordosis varied between 41° and 82°, and the number of vertebral bodies in a lordotic orientation varied from 1 to 8. The characteristics of the lumbar lordosis were most dependent on the orientation of the sacral slope and the pelvis. The upper arc of lumbar lordosis remained relatively constant, with an average value of approximately 20°. In contrast, the lower arc of lordosis was the most important determinant of the global lordosis. They concluded that changes in the specific sagittal alignment and lumbar lordosis are potentially responsible to indicate degenerative changes and symptomatic back pain, instead of utilizing the vague term of lordosis.

Torsion

Axial rotation of the spine involves torsion of the intervertebral discs and impaction of the facet joints. It is considered to be a possible mechanism of spinal damage and pain, especially

to the lumbar spine (Hadjipavlou et al., 1999). The assumed injury mechanism is that shear loading of the annulus and the damage of the facet joints and ligament structures may lead to the segmental damage. However, the axial torque was not considered a significant factor to contribute to the disc degeneration in previous studies. Because the facet joints limit the torsion to small range (1-2°), it does not appear to allow enough stress to lead to the disc damage (Adams & Hutton, 1981). Likewise, in an intact disc, the facet joints and posterior ligaments are known to protect the intervertebral disc from torsional loading. Because the axis of rotation of a lumbar vertebra passes through the posterior part of the vertebral body, all the posterior elements of the moving vertebra swing around the axis during the axial rotation (Cossette et al., 1971). Quantitative analysis revealed that the disc contributes 35% of the resistance to torsion, and the remaining 65% are being exerted by the posterior elements (Farfan et al., 1970). Another experimental study showed that the facet joints contribute 42 to 54% of the torsional stiffness in an FSU (Asano et al., 1992).

In contrast, a study by Liu et al. (1985) supported that cyclic torsional loads can lead to the failures in the disc, facet, laminae, and capsular ligaments. The anterior and posterior components are probably damaged or irritated by the axial torque of the lumbar spine. The initial torsional loading on the vertical axis are borne by the posterior elements (White & Panjabi, 1990).

Biomechanical Performance of Painful Spine

Patients with low back pain demonstrate a change in the mobility of the spine (Pearcy et al., 1985), deficits in reaction time, coordination (Luoto et al., 1996), and postural control with reduced velocity (Marras & Wongsam, 1986) when compared with healthy subjects. Differentiating the mechanical performance of pathologic spine with the normal variation

associated with gender, aging, and physical function will be the basic step to determine the treatment methods for patients with spinal pain. However, previous studies have been limited to identifying the behaviors of the spinal kinematics of low back pain patients, instead of approaching the various biomechanical performance of the spine.

Vogt et al. (2001) performed a kinematic study of lumbar spine during the treadmill walking in patients of chronic low back pain. They found significantly shorter stride times and stride-to-stride variability in all anatomic planes in these patients, as compared with healthy subjects. Decreased stride time, suggesting smaller steps of the patients, could be interpreted as a rigid or more cautious walking pattern and a protective way to reduce or avoid the pain. The increased between-subject variability could be interpreted as patients' individual adaptations and adjustments in walking behavior. These findings were interpreted as changed neuromuscular strategies to maintain an effective manner of locomotion which could be mediated by the altered proprioception in the lower back region. Nevertheless, the overall pattern of angular spinal displacements in patient group was shown to be within normal limits. It was suggested that pain of musculoskeletal origin had no significant effect on the magnitude of lumbar angular displacements.

Shum et al. (2005) performed a kinematic study of the lumbar spine and hip during sitting and standing movements in patients of low back pain. They found that the mobility and velocity of the spine and hip were significantly limited and the contribution of the lumbar spine relative to that of the hip was reduced in back pain subjects. The patients with low back pain, in particular those with positive straight leg raising (SLR) sign, had altered hip-spine coordination. Shum et al. (2006) performed another spine kinematic study for a picking up activity, and found the same reduced mobility of lumbar spine and hip in low back pain subjects.

Al-Eisa et al. (2006) completed a kinematic study of the trunk in patients of unilateral nonspecific low back pain during sitting. They found a significant correlation between pelvic asymmetry and asymmetric trunk motion in the patient group, and suggested that people with low back pain may have a distinct compensatory mechanism, secondary to the pelvic asymmetry from the unilateral low back pain, which put the lumbar spine under higher stresses. They concluded that movement asymmetry, rather than ROM, may be a better indicator of disturbed function for people with low back pain.

McGill et al. (1999) studied the motions of the spine and trunk muscle activity in normal elderly subjects (without back pain) during trunk posturing movements. They found the elderly exhibited slower motion, reduced ROM in full flexion and lateral bending. Furthermore, there was more coupled motion in the twisting efforts and abdominal muscles appeared to become more active earlier in the lateral bending movement. The earlier activation and increased co-contraction suggested that elderly people might be seeking greater stabilization either for general balance or for actual spine stabilization.

Landing Biomechanics

Previous studies on landing mechanics were mostly descriptive in biomechanical properties of various landing conditions. In majority of previous studies focused on defining injury mechanism (Fagenbaum & Darling, 2003; Wikstrom et al., 2006), gender differences (Kernozek et al., 2005), and conditional variability (Schot et al., 2002) of landing mechanics, associated with knee and ankle joints. There were several different techniques and models developed for the landing biomechanics of lower extremity joints (Baca, 1999; Nagano et al., 1998; Spagele et al., 1999). Additionally, there were several landing techniques tested to compare the efficiencies and injury rates during landing activities (McNair et al., 2000; Onate et al., 2005). Because landing from a variety of jumping is a good example of active movements,

the ability of a subject to control kinetic and kinematic changes applied to his/her body segments from landing can be a good measure to determine the functional status. However, most studies focused on the performance of lower extremity joints, and to our best knowledge none were developed to evaluate the performance of upper body segments above the hip joints during landing so far.

Biomechanical Performance of Lower Extremity Joints during Landing

Landing involved in many activities are vigorous and violent in nature. There are many reports about landing injuries from various activities (Ford et al., 2003; Salci et al., 2004). Many studies have been conducted to define the biomechanical etiology of lower extremity injuries during landing procedures (James et al., 2003; McNitt-Gray, 1993; Spagele et al., 1999). Vertical ground reaction force (GRF) from the ground impact of landing ranged from 1.0 to 14.0 times of body weight for normal subjects (Caulfield & Garrett, 2004; Decker et al., 2002; Fritz & Peikenkamp, 2001; James et al., 2003). Landing conditions investigated include landing height (McNitt-Gray, 1993; Zhang et al., 2000), landing posture and technique (Eloranta, 1996; Kovacs et al., 1999), joint stiffness of lower extremity joints (Devita & Skelly, 1992; Horita et al., 2002), and the number of legs involved in landing (Caulfield & Garrett, 2004; Hass et al., 2003).

Devita and Skelly (1992) studied biomechanical variables applied to the lower extremity joints from drop landing with a fall-height of 59 cm, comparing soft and stiff landing conditions. Soft and stiff landings were defined with knee flexion angle after ground impact as greater and less than 90°. The shapes of GRF, moment, and power curves were identical between both landings. Larger GRF, hip extensor, knee flexor, and ankle plantar flexor moments were observed during descent in the stiff landing, which produced a more erect body posture and a flexed knee position at impact. Also, the stiff landing exhibited larger power in ankle muscles,

while the soft landing showed larger powers in knee and hip muscles. They concluded that ankle plantar flexors absorbed more energy in the stiff landing, whereas the hip and knee extensors absorbed more energy in the soft landing.

To compare gymnasts and recreational athletes, McNitt-Gray (1993) evaluated lower extremity joints kinetics during drop landing with different landing heights. They found gymnasts chose to dissipate the impact loads by using the larger ankle and hip extensor moments at higher impact velocities than recreational athletes who chose to adjust their strategy by using greater degrees of hip flexion and longer landing phase durations than the gymnasts. The greater demands placed on the ankle and hip extensors by the gymnasts, as compared to the recreational athletes, was explained by the need to maintain balance during competitive gymnastics landings or, by the inability of recreational athletes to produce larger extensor moments at the ankle or hip during landings from the great heights.

Zhang et al. (2000) studied lower extremity joints kinetics with different landing heights (0.32 m, 0.62 m, and 1.03 m) and different landing techniques (soft, normal, and stiff landings). They found increases in peak GRF, peak joint moments, and powers with increases in landing height and stiffness. The mean eccentric work were 0.52, 0.74, and 0.87 J/kg by the ankle plantar flexors, 1.21, 1.63, and 2.26 J/kg by the knee extensors, and 0.94, 1.31, and 2.15 J/kg by the hip extensors, for heights of 0.32, 0.62, and 1.03 m, respectively. They concluded that knee extensors were consistent contributors to energy dissipation, while the ankle plantar flexors contributed more in the stiff landings and the hip extensors did more in the soft landings. This shift from ankle to hip strategy was observed as landing height increased. They explained that larger volume of proximal muscles (knee and hip joint muscles) of the lower extremity was more capable of energy absorption compared with the ankle muscle group. Another study suggested

that biarticular muscles are used effectively for power transportation during locomotion (Bobbert & van Ingen Schenau, 1988).

Gender Difference

The majority of knee injuries occurred during landing from various activities are caused by the non-contact mechanism. Numerous studies have found females have a higher rate of non-contact anterior cruciate ligament (ACL) injuries compared to males (Decker et al., 2003; Kernozek et al., 2005). Many studies had been conducted to identify different mechanical properties of landing in each gender and etiology of this gender disparity. Anatomically or intrinsically, small cross-sectional area of ACL (Feagin & Lambert, 1985), narrower intercondylar notch (Souryal et al., 1988), greater quadriceps angle, and greater knee laxity (Malinzak et al., 2001) of females have been suggested to contribute to the higher ACL injuries in females than in males. Except for the intrinsic anatomical factors, the differences in level of conditioning, level of muscle strength, and motor control strategies in females were suggested as the extrinsic factors, related with gender disparity (Delfico & Garrett, 1998).

Chappell et al. (2002) compared knee kinetics of male and female recreational athletes performing forward, vertical, and backward stop-jump tasks. They observed females exhibited greater proximal anterior shear force, greater knee extension and valgus moments than males did during the landing phase. During the takeoff phase, males showed greater proximal tibia anterior shear force than females. They concluded that females might have altered motor control strategies that result in knee positions in which ACL injuries might occur.

Decker et al. (2003) compared the biomechanical variables of lower extremity joints between male and female subjects performing a 60 cm drop landing. They found females demonstrated a more erect landing posture and utilized greater hip and ankle joint range of motions and maximum joint angular velocities when compared to males. Females exhibited

greater energy absorption and peak powers from the knee extensors and ankle plantar-flexors compared to males. Energy absorption contributions revealed that the knee extensor was the primary shock absorber for both genders. The ankle plantar flexor was the second largest contributor to energy absorption for the females and the hip extensor was for the males. The different shock absorption strategy used in females was proposed to provide a greater potential risk for non-contact ACL injury for females under certain landing conditions.

Kernozek et al. (2005) compared gender differences in frontal and sagittal plane biomechanics during a 60-cm drop landing. They observed that females exhibited greater peak hip and knee flexion, and ankle dorsiflexion angles in the sagittal plane, and greater peak knee valgus and ankle pronation angles in the frontal plane. Females exhibited greater peak vertical and posterior GRF, and reduced varus moment than males. They noted the importance of gender differences in the frontal plane variables in addition to those in sagittal plane.

Landing Stiffness

During landing from various activities, the actions of different musculoskeletal structures, including muscles, tendons, and ligaments, are integrated together so that the overall skeletons behave like a spring. As a result, landing from a jump can be modeled by using a simple spring-mass system (Asmussen & Bonde-Petersen, 1974). Stiffness of the leg spring represents the overall stiffness of the integrated musculoskeletal system during the landing phase, which is referred to as leg stiffness. Leg stiffness influences the kinetics and kinematics applied to the whole body during landing. Leg stiffness is greatly determined by the knee joint stiffness, which is mostly affected by knee extensor muscles. Joint stiffness is calculated by a linear regression of the knee joint moment/angle relationship. The interaction between leg stiffness and reflex activities plays a major role in regulating muscle power and performance in pre- and post-

landing to absorb or dissipate the large energy developed from the ground impact (Horita et al., 1996).

Horita et al. (2002) studied the interaction between pre-landing activities and stiffness regulation of the knee joint during a drop landing followed by a countermovement jump by examining landing motions, GRF, and EMG activity of the vastus lateralis during the pre- and post-landing. They divided the contact phase into three phases from initial contact to takeoff of the countermovement jump: (1) initial impact to initial peak of knee joint moment, (2) initial peak to onset of pushoff, and (3) concentric phase until takeoff. Drop landing performance was evaluated using the takeoff velocity, average contact time, and knee joint moment. A positive correlation was found between positive peak power of knee joint and the knee joint moment. However, they did not find any significant relationships between any drop landing performance parameters and ankle measures. They explained leg stiffness with a combination of pre-contraction of the vastus lateralis muscle and knee joint angular velocity at touchdown. They proposed two types of landing motions with regard to the pre-landing motion of the knee joint. The proper pre-landing movement could be characterized by the knee flexion just before touchdown, which is associated with a high initial joint stiffness coupled with the high joint power. This was called bouncing type in their study, which is close to the plyometrics. On the other hand, an inadequate pre-landing movement, associated with incomplete knee flexion induced subsequent deep-knee flexion after touchdown was called the absorbing type of landing and was regarded as the poor type. The absorbing movement comes too late and demand longer contact time and lower takeoff velocity.

The pre-activation of landing is initiated by the centrally pre-programmed motor commands of the required landing task (Dyhre-Poulsen et al., 1991), and increase the sensitivity

of the muscle spindle to enhance the stretch reflex (Gottlieb et al., 1981). The stretch reflex enhances muscle stiffness, and thus, improved stiffness regulation could be attained by proper pre-landing muscle activation (Allum & Mauritz, 1984).

GRF acting on the body during landing has been implicated in injuries to the lower extremity, and controlling peak GRF at impact is directly associated with lowering the landing stiffness. Apparently, the movement of lower extremity joints can influence the magnitude of the impact forces. It is generally known that subjects who land on the balls of their feet and flex their knee and ankle joints more have lower peak GRF. It was suggested that more joint movements allowed the body mass to decelerate over a longer period of time thus the impact force and time to peak force were decreased.

McNair et al. (2000) reported that GRF could be decreased immediately after instruction of landing technique about the lower extremity joints kinematics. Additionally, they commented that instruction could be more effective if the subjects' attention was drawn to distinct cues (sound of soft landing impact) in the environment.

Cowling et al. (2003) assessed the efficacy of verbal instruction about landing techniques to change impact force. They used instructions to increase knee flexion, to recruit hamstring muscles earlier, and muscle bursts immediately before landing. Only the instruction to increase knee flexion resulted in significantly greater knee flexion at initial ground contact and lower GRF, compared with the other instruction conditions.

Park et al. (2006a) compared mechanical characteristics of soft landing between male and female subjects performing vertical leaps and drop landings. They found significant increases in ankle plantar flexion at touchdown, knee flexion motion during the landing phase, times for maximal flexion of all joints, and decreases in peak vertical GRF, axial hip joint force, and knee

extensor moments in soft landing. Peak vertical GRF of males was significantly greater than that of females in soft landing. Pre-landing extension of the distal joints in vertical leap and that of proximal and distal joints in drop landing were suggested to activate the soft landing, with pre-contracting the muscles to prepare the proper landing. They noted that the soft landing strategy for males was fit for plyometrics and that of females was for absorbing type of landing.

Performance of Adapted Landing Biomechanics to Various Conditions

Landing is ideally suited for a performance study of weight-bearing segments of body, as it requires large eccentric muscle forces during the control of joint flexion and mimics the muscular stresses experienced during the landing phase. Therefore, numerous landing studies have been conducted to test the performance of normal subjects in various conditions, as well as to test subjects of ACL deficiency, ankle instability, fatigue of knee joint muscles, and so on.

A lesion of ACL is a major trauma of the knee joint, and mostly treated with a ligament reconstruction. ACL reconstruction demands sophisticated rehabilitation program to regain the original function before the injury. Decker et al. (2002) studied the biomechanics of lower extremity joints in fully rehabilitated ACL reconstructed and healthy subjects performing drop landing from a 60-cm height. At initial touchdown, the ACL group demonstrated greater hip extension and ankle plantar flexion, compared to the healthy group. The peak vertical GRF was not different between groups, but the ACL group delayed the time to its occurrence. The knee extensors provided the major energy absorption function for both groups; however, the ACL group performed 37% more ankle plantar flexor work and 39% less hip extensor work compared with the healthy group. They concluded that the ACL group utilized a different landing strategy adapted to the ACL reconstruction which employed the hip extensor muscles less and the ankle plantar flexors more.

Doorenbosch et al. (2003) studied EMG activity of the quadriceps and hamstrings in patients with ACL deficiency and healthy subjects during the vertical jump and landing. They observed significantly higher co-contraction index of quadriceps/hamstring muscles in ACL deficiency subjects. This was suggested that higher levels of co-contraction of quadriceps and hamstrings during movements in ACL deficient subject help to compensate for the loss of the passive constraining structure.

Caulfield et al. (2004) studied jump and landing performance in subjects with functional instability of the ankle joint and normal control. The subjects performed five single leg jumps onto a forceplate. They observed lateral and anterior force peaks occurred significantly earlier in subjects with ankle instability. These changes occurred immediately post-impact and too early for the reflex correction.

Madigan et al. (2003) studied the effect of lower extremity fatigue on the performance of lower extremity joints during drop landing. EMG data were used to confirm fatigue in the quadriceps muscles. They observed a decrease in peak vertical GRF, an increase in maximum flexion of knee and ankle joints occurred early in a fatigue landing, while significant changes in vertical GRF impulse and time to maximum knee flexion occurred during the middle or late stages of a fatigue landing. For the first half of a fatigue landing, hip extensor impulse increased, knee extensor impulse did not change, and ankle plantar flexor impulse decreased. These changes were explained with a distal-to-proximal redistribution of extensor moments, which allowed the larger proximal muscles to contribute more to resisting collapse during landing. They also suggested active insufficiency of gastrocnemius, since it crosses the knee and shortens as knee flexion increases. The shortening of this muscle diminishes the ability to produce plantar flexor moment at the ankle. The increase in knee flexion upon landing during the first half of the

fatigue landing may have reduced the contribution of the gastrocnemius muscle to plantar flexor impulse and resulted in an overall decrease in plantar flexor impulse. This decrease also has dictated an increase in hip extensor impulse to generate the necessary support moment for landing deceleration.

Park et al. (2006b) studied the effects of limited lower back motion on soft landing mechanics of lower extremity joints with subjects wearing various low back braces, simulating different low back stiffness conditions. They found that knee and hip joint flexions were decreased and peak vertical GRF and axial hip force were increased in stiff brace condition, compared with no brace and soft brace conditions. Additionally, typical sequential joint flexion from distal to proximal was disrupted in females wearing the stiff brace, comparing to male counterpart. They concluded that limited spinal motions by the brace caused alterations in knee and hip joint motions during the landing phase and an increase in impact force. They emphasized that lower back motion is one of the factors in determining landing mechanics, and a stiff lower back is associated with a stiff landing. With limited trunk motion, more decelerating torque might be concentrated on the knee extensors for females, and more axial loads be transmitted to the proximal body segments of males during the soft landing.

Table 3-1. Average neutral zones for a functional spinal units in different regions of the spine (°).

Region	Flexion/Extension (total)	Lateral bending (one side)	Axial rotation (one side)
C0 - C1	1.1	1.6	1.5
C1 - C2	3.2	1.2	29.6
C3 - C6	4.9	4	3.8
C7 - T1 / T11 - T1	1.5	2.2	1.2
L1 - L2 / L3 - L4	1.5	1.6	0.7
L5 - S1	3	1.8	0.4

Note: Cited from the work of White and Panjabi (1990).

Table 3-2. Representative ranges of motion of C0-C1-C2 complex (°).

Level	Reference	Flexion/Extension (total)	Lateral bending (one side)	Axial rotation (one side)
C0-C1	Penning (1978)	35	10	0
	Goel et al. (1988)	23	3.4	2.4
	Panjabi et al. (1988)	24.5	5.5	7.2
C1-C2	Penning (1978)	30	10	70
	Goel et al. (1988)	10.1	42	23.3
	Panjabi et al. (1988)	22.4	6.7	38.9

Table 3-3. Representative ranges and limits of motion of the middle and lower cervical spines (°).

Region	Flexion/Extention	Lateral bending (one side)	Axial rotation (one side)
Middle			
C2-C3	10 (5 - 16)	10 (11 - 20)	3 (0 - 10)
C3-C4	15 (7 - 26)	11 (9 - 15)	7 (3 - 10)
C4-C5	20 (13 - 29)	11 (0 - 16)	7 (1 - 12)
Lower			
C5-C6	20 (13 - 29)	8 (0 - 16)	7 (2 - 12)
C6-C7	17 (6 - 26)	7 (0 - 17)	6 (2 - 10)
C7-T1	9 (4 - 7)	4 (0 - 17)	2 (0 - 7)

Note: Cited from the work of White and Panjabi (1990).

Table 3-4. Normal active cervical ranges of motion (*in vivo*) reported in the literatures (°).

Motion	Dvorak (1992)	Lanz (1999)	AMA (2001)
Flexion/Extension	141.3	116.1	110
Lateral bending	91.4	84.1	90
Axial rotation	175	144.2	160
Rotation from flexion	81.4	–	–
Rotation from extension	165	–	–

Table 3-5. Representative ranges and limits of motion of the thoracic spine (°).

Level	Flexion/Extension (total)	Lateral bending (one side)	Axial rotation (one side)
T1 - T2	4 (3 – 5)	5 (5)	9 (14)
T2 - T3	4 (3 – 5)	6 (5 – 7)	8 (4 – 12)
T3 - T4	4 (2 – 5)	5 (3 – 7)	8 (5 – 11)
T4 - T5	4 (2 – 5)	6 (5 – 6)	8 (5 – 11)
T5 - T6	4 (3 – 5)	6 (5 – 6)	8 (5 – 11)
T6 - T7	5 (2 – 7)	6 (6)	7 (4 – 11)
T7 - T8	6 (3 – 8)	6 (3 – 8)	7 (4 – 11)
T8 - T9	6 (3 – 8)	6 (4 – 7)	6 (6 – 7)
T9 - T10	6 (3 – 8)	6 (4 – 7)	4 (3 – 5)
T10 - T11	9 (4 – 14)	7 (3 – 10)	2 (2 – 3)
T11 - T12	12 (6 – 20)	9 (4 – 13)	2 (2 – 3)
T12 - L1	12 (6 – 20)	8 (5 – 10)	2 (2 – 3)

Note: Cited from the work of White and Panjabi (1990).

Table 3-6. Representative ranges and limits of motion of the lumbar spine (°).

Level	Flexion/Extension	Lateral bending	Axial rotation
L1 - L2	12 (5 - 16)	6 (3 - 8)	2 (1 - 3)
L2 - L3	14 (8 - 18)	6 (3 - 10)	2 (1 - 3)
L3 - L4	15 (6 - 17)	8 (4 - 12)	2 (1 - 3)
L4 - L5	16 (9 - 21)	6 (3 - 9)	2 (1 - 3)
L5 - S1	17 (10 - 24)	3 (2 - 6)	1 (0 - 2)

Note: Cited from the work of White and Panjabi (1990).

Table 3-7. Comparison of lumbar compression loads in various trunk postures without external loading.

	Nachemson (1966; 1981)		Takahashi et al. (2006)		Ledet et al. (2005)	
Trunk posture	N	% of standing	N	% of standing	× body weight	% of standing
Supine	300	25	–	–	1.9	95
Standing	700	100	645	100	2	100
Sitting	1000	140	–	–	2.5	125
Standing flexed	–	–	–	–	2.6*	130*
10°	–	–	1277	198	–	–
20°	1200	150	1922	298	–	–
30°	–	–	2305	357	–	–
Sitting flexed	–	185	–	–	2.8	140

* The trunk flexion angles were not specified in Ledet et al. (2005).

Table 3-8. Average ranges of motion of the lumbar spine in normal walking and running in different studies (°).

Source	Flexion/Extension	Lateral bending	Axial rotation
Walking			
Crosbie et al. (1997)	3.5	9	4.5
(Lower thoracic)	2.5	7.0	4.0
(Pelvis)	3.5	6.0	4.0
Callaghan et al. (1999)	6.2	6.7	7.1
Van Herp et al. (2000)	2.3	4	6.6
Running			
Schache et al. (2002)	13.3	18.5	23.0
(Pelvis)	7.6	10.6	13.9

Table 3-9. Peak compression loads to the lower lumbar level during walking (× BW).

Source	Model	Compression force
Cappozzo (1983)	Single muscle equivalent model	1.0 - 2.5
Cromwell et al. (1989)	EMG model	1.0
Khoo et al. (1995)	Single muscle equivalent model	1.5 - 2.1
Callaghan et al. (1999)	EMG model	0.9 - 3.5
	Inverse dynamic model	0.2 - 1.0
Nachemson (1964)	Direct intradiscal pressure measurement	850 (N)

Table 3-10. Intradiscal pressure of low lumbar level during various activities.

	Wilke et al. (1999)	Rohlmann et al. (2001)	Nachemson (1966; 1981; 1987)
Position	Intradiscal pressure (MPa)	Percentage of standing (%)	Percentage of standing (%)
Lying supine	0.1	20	25
Relaxed standing	0.5	100	100
Standing with forward bending	1.1	216	150
Sitting without backrest	0.46	90	140
Sitting with maximum flexion	0.83	–	185
Standing up from a chair	1.1	–	–
Walking	0.53 - 0.65	130	121
Jogging	0.35 - 0.65	170	–
Jumping	–	240 - 380	157
Lifting 20 kg, squat lift	1.7	–	300
Lifting 20 kg, stoop lift	2.3	460	486

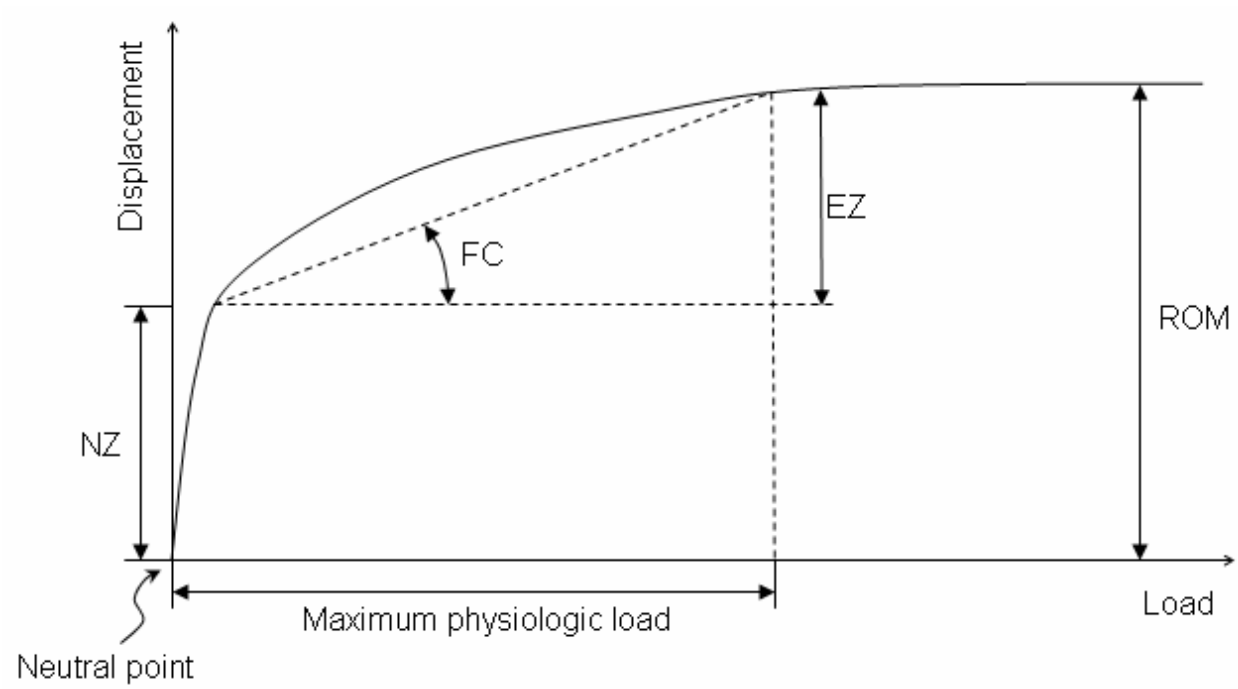

Figure 3-1. The load-displacement curve of a functional spinal unit (FSU) is generally nonlinear and biphasic [neutral zone (NZ) and elastic zone (EZ)]. Range of motion (ROM) is the sum of the neutral and elastic zones. Average flexibility coefficient (FC) is the elastic zone divided by the maximum physiological load.

CHAPTER 4
RESULTS

Effects of Landing Technique

Overall landing characteristics were evaluated using different landing variables (Table 4-1). Repeated measures MANOVA revealed significant main effects of gender ($p<0.001$) and landing technique ($p<0.001$) for the landing variables. However, no significant interaction was found between gender and landing technique ($p=0.057$) (Table C-1 in Appendix C).

Follow-up univariate contrast procedures revealed that $t_{(LP)}$, $\theta_{P(KFA)}$, and $\theta_{P(HFA)}$ increased significantly and PVGRF decreased significantly when going from NL to SL condition. For touchdown angles, both $\theta_{TD(KFA)}$ and $\theta_{TD(HFA)}$ were significantly more flexed when going from NL to SL condition. Female subjects exhibited significantly greater $\theta_{P(HFA)}$ than male subjects across both landing conditions (Table 4-1).

Kinematic characteristics of the spinal column after touchdown of drop landing were evaluated with touchdown angle and extension ROM variables of each spinal region. Relative to research questions Q1 and Q2 and associated hypotheses (1a and 2a), repeated measures MANOVA revealed a significant main effect of gender ($p=0.011$) and a significant interaction between gender and landing technique ($p=0.025$) for touchdown angle variables, and significant main effects of gender ($p=0.013$) and landing technique ($p<0.001$) for extension ROM variables. However, no significant main effect of landing technique was found for touchdown angle variables ($p=0.236$), and no significant interaction was found between gender and landing technique for extension ROM variables ($p=0.232$) (Table C-1).

Follow-up univariate contrast procedures for touchdown angle variables revealed that $\gamma_{TD(TL/L)}$ of females was significantly greater than that of males (i.e., females demonstrated significantly more extended thoracolumbar regional angle at touchdown than males), and a

significant interaction was found in $\gamma_{TD(TH/TL)}$ (Table 4-2). The significant interaction of $\gamma_{TD(TH/TL)}$ between gender and landing technique means that males demonstrated more flexed thoracic regional angle at touchdown from NL to SL condition, while females did not demonstrate differences across different landing techniques (Figure 4-1).

Follow-up univariate tests for extension ROM variables identified a significant main effect of gender for $t_{P(TH/TL)}$ and significant main effects of landing technique for $\gamma_{P(TH/TL)}$ and $\gamma_{P(LC/TH)}$. The contrast procedures indicated that females exhibited significantly greater extension motion in the thoracic region than males did, and extension motions in the thoracic and lower cervical regions increased significantly from NL to SL condition (Table 4-2).

The overall kinematic characteristics of each spinal region reveal that the lumbar and thoracolumbar regions exhibit flexion and the thoracic and lower cervical regions show extension during the landing phase. However, the thoracolumbar region undergoes a short period of extension followed by flexion during the landing phase in selected subjects (Figure 4-2).

Kinetic characteristics of L/S and C/T junctions after touchdown during drop landing were evaluated with the kinetic variables at L/S and C/T junctions. Relative to research questions Q1 and Q2 and associated hypotheses (1b and 2b), repeated measures MANOVA revealed a significant main effect of landing technique for the kinetic variables at L/S junction ($p<0.001$), and significant main effects of gender ($p=0.031$) and landing technique ($p<0.001$) for the kinetic variables at C/T junction. There was also a significant interaction between gender and landing technique for the kinetic variables at C/T junction ($p=0.018$). However, no significant gender effect ($p=0.403$) or interaction ($p=0.196$) between gender and landing technique was found for the kinetic variables at L/S junction (Table C-1). The univariate contrast procedures revealed that

all the kinetic variables at L/S junction decreased significantly from NL to SL condition (Table 4-3).

For the kinetic variables at C/T junction, all but $ShF(C/T)_{ant}$ decreased significantly from NL to SL condition and $ShF(C/T)_{ant}$ and $ShF(C/T)_{post}$ demonstrated significant interactions between gender and landing technique (Table 4-3). The significant interactions between gender and landing technique for $ShF(C/T)_{ant}$ and $ShF(C/T)_{post}$ indicate that $ShF(C/T)_{post}$ of males was greater than that of females during NL, but no gender difference was observed during SL condition. On the other hand, $ShF(C/T)_{ant}$ of females was greater than that of males during NL, but not different from each other during SL condition. Females demonstrated decreased $ShF(C/T)_{ant}$ from NL to SL condition, but not for males (Figure 4-1).

Effects of Knee Joint Muscles Fatigue

Overall landing characteristics were evaluated using different landing variables. Repeated measures MANOVA identified a significant main effect of gender (p=0.012) for the landing variables. However, no significant fatigue effect (p=0.559) or interaction (p=0.104) between gender and fatigue level was found for the landing variables (Table C-9). Follow-up univariate tests revealed that female subjects exhibited a significantly greater $\theta_{TD(KFA)}$ than male subjects across both landing conditions (i.e., females demonstrated more extended posture of knee joint at touchdown than males) (Table 4-4).

Kinematic characteristics of the spinal column after touchdown of drop landing were evaluated with touchdown angle and extension ROM variables. Relative to research questions Q3 and Q4 and associated hypotheses (3a and 4a), repeated measures MANOVA revealed significant main effects of gender for the touchdown angle (p=0.036) and extension ROM

(p=0.006) variables. However, there was not any significant fatigue effect or interaction for both touchdown angle and extension ROM variables (Table C-9).

Univariate contrast procedures revealed that females demonstrated a significantly extended posture of thoracolumbar region at touchdown (greater $\gamma_{TD(TL/L)}$ in females than in males) and a greater extension motion (greater $\gamma_{P(TH/TL)}$ in females than in males) during landing phase than males did (Table 4-5).

The overall kinematic characteristics of each spinal region reveal that the lumbar region exhibits flexion and thoracic and lower cervical regions show extension during the landing phase. However, the thoracolumbar region undergoes a short period of extension followed with flexion during the landing phase in selected subjects (Figure 4-3).

Kinetic characteristics of L/S and C/T junctions after touchdown of drop landing were evaluated with the kinetic variables of L/S and C/T junctions. Relative to research questions Q3 and Q4 and associated hypotheses (3b and 4b), repeated measures MANOVA revealed significant interactions between gender and fatigue level for the kinetic variables at L/S junction (p=0.033) and C/T junction (p=0.043) (Table C-9).

Follow-up univariate tests identified significant interactions of gender × fatigue level for AxF(L/S), $ShF(L/S)_{post}$, ExtM(L/S), and AxF(C/T) (Table 4-6). The significant interaction for AxF(L/S) indicated that females exhibited increased AxF(L/S) from SL1 to SL2 condition, while males did not (Figure 4-4). The significant interaction of $ShF(L/S)_{post}$ revealed that females exhibited increased $ShF(L/S)_{post}$ from SL1 to SL2 condition, while males did not. Also, $ShF(L/S)_{post}$ was greater in males than in females for SL1, but was greater in females than in males for SL2 condition. The significant interaction of AxF(C/T) indicated that females exhibited increased AxF(C/T) from SL1 to SL2 condition, while males did not. Also, AxF(C/T)

was greater in males than in females for SL1, but was greater in females than in males for SL2 condition. The significant interaction of ExtM(L/S) revealed that females exhibited increased ExtM(L/S) from SL1 to SL2 condition, while males did not. Lastly, ExtM(L/S) was greater in males than in females for SL1, but was greater in females than in males for SL2 condition.

Table 4-1. Collapsed mean and SD values of different landing variables for different genders and landing techniques.

Landing variables	NL Mean (SD)	SL Mean (SD)	Gender: p	Landing: p	Landing × Gender: p
PVGRF ($N \cdot kg^{-1}$)	23.91 (3.87)	15.77 (3.03)	0.877	<0.001*	0.723
Male	23.67 (3.22)	15.83 (2.91)			
Female	24.15 (4.55)	15.70 (3.27)			
$t_{(LP)}$ (s)	0.222 (0.084)	0.355 (0.139)	0.148	<0.001*	0.536
Male	0.203 (0.071)	0.320 (0.075)			
Female	0.241 (0.095)	0.390 (0.179)			
$\theta_{TD(KFA)}$ (°)	-7.3 (6.8)	-10.5 (6.4)	0.063	<0.001*	0.268
Male	-9.2 (4.6)	-13.2 (5.5)			
Female	-5.3 (8.2)	-7.8 (6.2)			
$\theta_{P(KFA)}$ (°)	53.0 (10.4)	69.9 (14.2)	0.087	<0.001*	0.131
Male	51.7 (12.6)	71.8 (16.0)			
Female	54.2 (7.9)	68.0 (12.6)			
$\theta_{TD(HFA)}$ (°)	-61.5 (12.0)	-66.6 (12.3)	0.853	<0.001*	0.747
Male	-60.9 (10.7)	-66.4 (11.7)			
Female	-62.1 (13.6)	-66.9 (13.4)			
$\theta_{P(HFA)}$ (°)	33.3 (13.7)	51.2 (10.8)	0.001*	<0.001*	0.113
Male	24.8 (11.5)	47.0 (7.2)			
Female	41.7 (10.2)	55.5 (12.3)			

Note: NL (self-selected normal landing), SL (soft landing), PVGRF (peak vertical GRF), $t_{(LP)}$ (time for landing phase), $\theta_{TD(KFA)}$ (knee flexion angle at touchdown),$\theta_{P(KFA)}$ (ROM of knee flexion angle from touchdown to initial peak of knee flexion), $\theta_{TD(HFA)}$ (hip flexion angle at touchdown), $\theta_{P(KFA)}$ (ROM of hip flexion angle from touchdown to initial peak of hip flexion), * (significant in univariate tests)

Table 4-2. Collapsed mean and SD values of touchdown angle and extension ROM of each spinal region for different genders and landing techniques.

Kinematic variables	NL Mean (SD)	SL Mean (SD)	Gender: p	Landing: p	Landing × Gender: p
Touchdown angle of each spinal region					
$\gamma_{TD(L/S)}$ (°)	15.1 (8.8)	15.0 (9.0)	0.215	0.85	0.433
Male	13.3 (9.0)	12.6 (9.5)			
Female	17.0 (8.5)	17.4 (8.2)			
$\gamma_{TD(TL/L)}$ (°)	11.9 (7.8)	11.9 (8.7)	0.003*	0.933	0.28
Male	7.6 (5.1)	7.2 (6.1)			
Female	16.1 (7.8)	16.7 (8.4)			
$\gamma_{TD(TH/TL)}$ (°)	-21.9 (6.9)	-22.6 (7.8)	0.196	0.219	0.032*
Male	-23.1 (6.2)	-25.1 (8.2)			
Female	-20.7 (7.6)	-20.1 (6.9)			
$\gamma_{TD(LC/TH)}$ (°)	-1.6 (12.4)	-0.3 (13.4)	0.91	0.451	0.844
Male	-1.4 (10.6)	0.1 (14.6)			
Female	-1.7 (14.5)	-0.8 (12.8)			
Extension ROM of each spinal region					
$\gamma_{P(L/S)}$ (°)	0.5 (1.3)	0.0 (0.1)	0.167	0.061	0.179
Male	0.2 (0.3)	0.0 (0.1)			
Female	0.9 (1.7)	0.1 (0.1)			
$\gamma_{P(TL/L)}$ (°)	2.3 (2.0)	2.0 (2.3)	0.070	0.392	0.526
Male	1.8 (1.9)	1.2 (1.7)			
Female	2.9 (2.0)	2.8 (2.6)			
$\gamma_{P(TH/TL)}$ (°)	4.6 (3.7)	8.0 (5.1)	0.012*	< 0.001*	0.057
Male	3.3 (2.8)	5.4 (3.9)			
Female	6.0 (4.2)	10.6 (4.9)			
$\gamma_{P(LC/TH)}$ (°)	10.3 (7.8)	16.3 (10.4)	0.591	0.006*	0.557
Male	10.0 (9.7)	14.8 (10.8)			
Female	10.5 (5.7)	17.7 (10.1)			

Note: NL (self-selected normal landing), SL (soft landing), $\gamma_{TD(L/S)}$ (lumbar regional angle at touchdown), $\gamma_{TD(TL/L)}$ (thoracolumbar regional angle at touchdown), $\gamma_{TD(TH/TL)}$ (thoracic regional angle at touchdown), $\gamma_{TD(LC/TH)}$ (lower cervical regional angle at touchdown), $\gamma_{P(L/S)}$ (extension ROM of lumbar region from touchdown to initial peak during landing phase), $\gamma_{P(TL/L)}$ (extension ROM of thoracolumbar region from touchdown to initial peak during landing phase), $\gamma_{P(TH/TL)}$ (extension ROM of thoracic region from touchdown to initial peak during landing phase), $\gamma_{P(LC/TH)}$ (extension ROM of lower cervical region from touchdown to initial peak during landing phase), * (significant in univariate tests)

Table 4-3. Collapsed mean and SD values of kinetic variables at L/S and C/T junctions for different genders and landing techniques.

Kinetic variables	NL Mean (SD)	SL Mean (SD)	Gender: p	Landing: p	Landing × Gender: p
L/S junction					
AxF(L/S) ($N \cdot kg^{-1}$)	8.52 (2.70)	5.77 (1.93)	0.182	<0.001*	0.268
Male	7.72 (2.43)	5.55 (1.77)			
Female	9.33 (2.81)	5.99 (2.13)			
$ShF(L/S)_{ant}$ ($N \cdot kg^{-1}$)	1.73 (1.01)	0.83 (0.51)	0.085	<0.001*	0.022
Male	2.16 (1.24)	0.82 (0.55)			
Female	1.30 (0.45)	0.84 (0.48)			
$ShF(L/S)_{post}$ ($N \cdot kg^{-1}$)	9.73 (3.54)	4.35 (2.32)	0.297	<0.001*	0.476
Male	10.48 (3.21)	4.66 (2.45)			
Female	8.97 (3.81)	4.05 (2.23)			
FlxM(L/S) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	1.37 (0.44)	0.91 (0.43)	0.583	<0.001*	0.961
M	1.41 (0.37)	0.95 (0.36)			
F	1.33 (0.50)	0.87 (0.51)			
ExtM(L/S) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	3.44 (1.15)	1.94 (0.70)	0.31	<0.001*	0.418
Male	3.68 (1.06)	2.02 (0.64)			
Female	3.19 (1.23)	1.85 (0.77)			
C/T junction					
AxF(C/T) ($N \cdot kg^{-1}$)	4.98 (4.07)	1.65 (2.28)	0.749	0.001*	0.303
Male	4.36 (4.0)	1.96 (2.25)			
Female	5.60 (4.20)	1.34 (2.36)			
$ShF(C/T)_{ant}$ ($N \cdot kg^{-1}$)	3.22 (2.28)	2.56 (1.73)	0.197	0.117	0.042*
Male	2.35 (1.84)	2.56 (1.58)			
Female	4.08 (2.42)	2.56 (1.94)			
$ShF(C/T)_{post}$ ($N \cdot kg^{-1}$)	5.54 (1.28)	4.26 (0.81)	0.026*	<0.001*	0.001*
Male	6.26 (1.42)	4.35 (0.97)			
Female	4.83 (0.53)	4.17 (0.65)			
FlxM(C/T) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	2.54 (0.91)	1.75 (0.81)	0.758	<0.001*	0.342
Male	2.40 (0.77)	1.80 (0.70)			
Female	2.68 (1.04)	1.70 (0.93)			
ExtM(C/T) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	3.85 (1.33)	2.24 (0.71)	0.302	<0.001*	0.145
Male	4.21 (1.4	2.25 (0.60)			
Female	3.50 (1.19)	2.24 (0.83)			

Note: NL (self-selected normal landing), SL (soft landing), AxF (peak axial compressive force), $ShF_{ant(or\ post)}$ (peak ant. or post. shear force), FlxM (peak flexor moment), ExtM (peak extensor moment), (L/S) (for lumbosacral junction), (C/T) (for cervicothoracic junction), * (significant in univariate tests)

Table 4-4. Collapsed mean and SD values of different landing variables for different genders and fatigue levels.

Landing variables	SL1 Mean (SD)	SL2 Mean (SD)	Gender: p	Fatigue: p	Fatigue × Gender: p
PVGRF ($N \cdot kg^{-1}$)	15.72 (3.03)	16.47 (3.43)	0.312	0.087	0.007
Male	15.8 (3.04)	15.2 (3.36)			
Female	15.62 (3.19)	17.99 (2.98)			
$t_{(LP)}$ (s)	0.335 (0.147)	0.342 (0.163)	0.304	0.628	0.666
Male	0.307 (0.088)	0.308 (0.093)			
Female	0.368 (0.197)	0.382 (0.220)			
$\theta_{TD(KFA)}$ (°)	-10.2 (6.7)	-11.0 (7.1)	0.008*	0.322	0.281
Male	-13.2 (5.8)	-14.6 (6.5)			
Female	-6.6 (6.2)	-6.6 (5.2)			
$\theta_{P(KFA)}$ (°)	70.5 (15.0)	71.7 (15.9)	0.271	0.44	0.112
Male	72.8 (16.2)	75.9 (18.0)			
Female	67.6 (13.8)	66.5 (11.9)			
$\theta_{TD(HFA)}$ (°)	-65.2 (10.9)	-64.1 (15.5)	0.094	0.608	0.648
Male	-66.3 (11.8)	-64.2 (18.4)			
Female	-64.0 (10.2)	-63.9 (12.2)			
$\theta_{P(HFA)}$ (°)	52.0 (10.8)	53.6 (12.9)	0.094	0.264	0.749
Male	48.1 (6.3)	50.0 (8.5)			
Female	56.8 (13.3)	57.9 (16.3)			

Note: SL1 (soft landing before the fatigue procedures), SL2 (soft landing after the fatigue procedures), PVGRF (peak vertical GRF), $t_{(LP)}$ (time for landing phase), $\theta_{TD(KFA)}$ (knee flexion angle at touchdown),$\theta_{P(KFA)}$ (ROM of knee flexion angle from touchdown to initial peak of knee flexion), $\theta_{TD(HFA)}$ (hip flexion angle at touchdown), $\theta_{P(KFA)}$ (ROM of hip flexion angle from touchdown to initial peak of hip flexion), * (significant in univariate tests)

Table 4-5. Collapsed mean and SD values of touchdown angle and extension ROM of each spinal region for different genders and fatigue levels.

Kinematic variables	SL1 Mean (SD)	SL2 Mean (SD)	Gender: p	Fatigue: p	Fatigue × Gender: p
Touchdown angle of each spinal region					
$\gamma_{TD(L/S)}$ (°)	15.1 (9.0)	14.0 (9.3)	0.473	0.343	0.558
Male	13.5 (10.0)	13.1 (9.8)			
Female	17.0 (7.7)	15.1 (9.1)			
$\gamma_{TD(TL/L)}$ (°)	11.5 (9.0)	11.4 (9.9)	0.005*	0.988	0.545
Male	6.8 (6.4)	6.5 (7.0)			
Female	17.0 (8.7)	17.3 (9.8)			
$\gamma_{TD(TH/TL)}$ (°)	-22.9 (8.1)	-22.9 (7.8)	0.068	0.924	0.993
Male	-25.7 (8.2)	-25.7 (7.0)			
Female	-19.6 (7.1)	-19.6 (7.6)			
$\gamma_{TD(LC/TH)}$ (°)	0.3 (13.6)	3.2 (13.4)	0.819	0.056	0.184
Male	0.7 (15.5)	1.6 (15.2)			
Female	-0.1 (11.6)	5.0 (11.5)			
Extension ROM of each spinal region					
$\gamma_{P(L/S)}$ (°)	0.0 (0.1)	0.4 (1.2)	0.074	0.072	0.087
Male	0.0 (0.0)	0.0 (0.0)			
Female	0.0 (0.1)	0.9 (1.7)			
$\gamma_{P(TL/L)}$ (°)	2.1 (2.4)	1.5 (1.5)	0.085	0.083	0.212
Male	1.3 (1.8)	1.1 (1.2)			
Female	3.1 (2.7)	1.9 (1.6)			
$\gamma_{P(TH/TL)}$ (°)	7.8 (5.5)	8.1 (5.5)	0.013*	0.624	0.624
Male	5.2 (4.0)	5.7 (4.4)			
Female	11.0 (5.5)	11.0 (5.5)			
$\gamma_{P(LC/TH)}$ (°)	16.6 (10.6)	19.2 (12.3)	0.776	0.355	0.042
Male	15.0 (11.2)	22.0 (14.0)			
Female	18.6 (10.1)	15.9 (9.5)			

Note: SL1 (soft landing before the fatigue procedures), SL2 (soft landing after the fatigue procedures), $\gamma_{TD(L/S)}$ (lumbar regional angle at touchdown), $\gamma_{TD(TL/L)}$ (thoracolumbar regional angle at touchdown), $\gamma_{TD(TH/TL)}$ (thoracic regional angle at touchdown), $\gamma_{TD(LC/TH)}$ (lower cervical regional angle at touchdown), $\gamma_{P(L/S)}$ (extension ROM of lumbar region from touchdown to initial peak during landing phase), $\gamma_{P(TL/L)}$ (extension ROM of thoracolumbar region from touchdown to initial peak during landing phase), $\gamma_{P(TH/TL)}$ (extension ROM of thoracic region from touchdown to initial peak during landing phase), $\gamma_{P(LC/TH)}$ (extension ROM of lower cervical region from touchdown to initial peak during landing phase), * (significant in univariate tests)

Table 4-6. Collapsed mean and SD values of kinetic variables at L/S and C/T junctions for different genders and fatigue levels.

Kinetic variables	SL1 Mean (SD)	SL2 Mean (SD)	Gender: p	Fatigue: p	Fatigue × Gender: p
L/S junction					
AxF(L/S) ($N \cdot kg^{-1}$)	5.79 (2.01)	5.76 (1.86)	0.063	0.9	0.019*
Male	5.46 (1.82)	4.80 (1.38)			
Female	6.19 (2.26)	6.91 (1.73)			
$ShF(L/S)_{ant}$ ($N \cdot kg^{-1}$)	0.84 (0.53)	0.85 (0.41)	0.763	0.877	0.185
Male	0.88 (0.53)	0.76 (0.40)			
Female	0.80 (0.54)	0.95 (0.43)			
$ShF(L/S)_{post}$ ($N \cdot kg^{-1}$)	4.40 (2.48)	5.33 (2.45)	0.703	0.002	0.001*
Male	4.71 (2.55)	4.66 (2.39)			
Female	4.02 (2.46)	6.13 (2.39)			
FlxM(L/S) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	0.93 (0.45)	0.98 (0.42)	0.709	0.576	0.961
Male	0.96 (0.37)	1.01 (0.40)			
Female	0.90 (0.55)	0.94 (0.47)			
ExtM(L/S) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	1.93 (0.73)	2.21 (0.68)	0.7	0.01	0.011*
Male	2.02 (0.67)	2.02 (0.63)			
Female	1.82 (0.83)	2.44 (0.70)			
C/T junction					
AxF(C/T) ($N \cdot kg^{-1}$)	1.74 (2.40)	2.21 (2.04)	0.541	0.096	0.002*
Male	2.04 (2.33)	1.42 (1.28)			
Female	1.38 (2.56)	3.17 (2.42)			
$ShF(C/T)_{ant}$ ($N \cdot kg^{-1}$)	2.71 (1.71)	2.66 (1.72)	0.398	0.907	0.715
Male	2.51 (1.64)	2.35 (1.48)			
Female	2.95 (1.84)	3.04 (1.98)			
$ShF(C/T)_{post}$ ($N \cdot kg^{-1}$)	4.19 (0.86)	4.53 (0.90)	0.438	0.036	0.699
Male	4.28 (0.98)	4.69 (0.93)			
Female	4.07 (0.71)	4.35 (0.88)			
FlxM(C/T) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	1.84 (0.80)	1.94 (0.76)	0.782	0.474	0.787
Male	1.78 (0.73)	1.91 (0.63)			
Female	1.90 (0.92)	1.97 (0.92)			
ExtM(C/T) ($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	2.21 (0.72)	2.53 (0.70)	0.454	0.022	0.054
Male	2.24 (0.63)	2.30 (0.59)			
Female	2.17 (0.86)	2.79 (0.77)			

Note: SL1 (soft landing before the fatigue procedures), SL2 (soft landing after the fatigue procedures), AxF (peak axial compressive force), $ShF_{ant(or\ post)}$ (peak ant. or post. shear force), FlxM (peak flexor moment), ExtM (peak extensor moment), (L/S) (for lumbosacral junction), (C/T) (for cervicothoracic junction), * (significant in univariate tests)

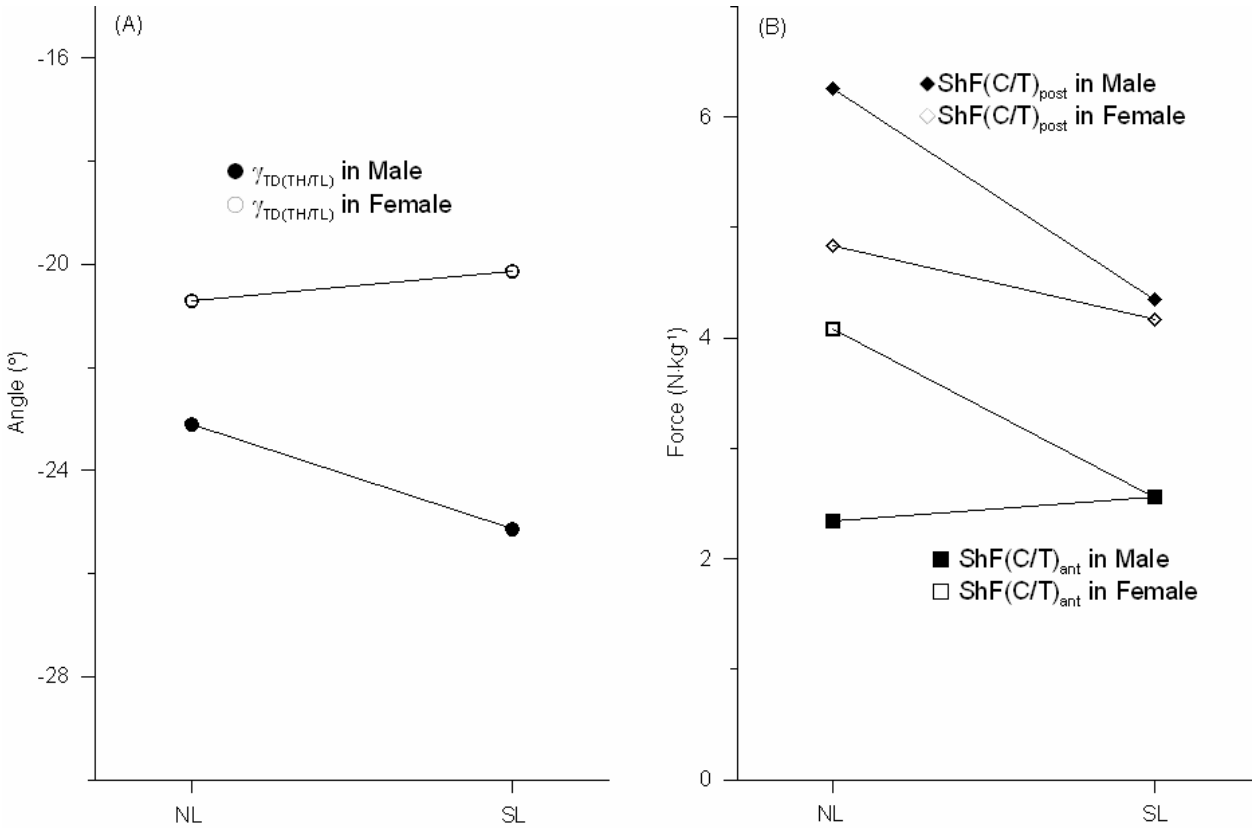

Figure 4-1. Significant interactions of the touchdown angle and C/T kinetic variables between gender and landing technique: (A) $\gamma_{TD(TH/TL)}$ was decreased from NL to SL condition in males, but not decreased in females (thoracic regional angle becomes more flexed from NL to SL condition in males, but not in females), (B) $ShF(C/T)_{post}$ of males was greater than that of females during NL, but not different from each other during SL condition; $ShF(C/T)_{ant}$ of females was greater than that of males during NL, but not different from each other during SL condition; $ShF(C/T)_{ant}$ was decreased from NL to SL condition in females, but not decreased in males. Note: NL (self-selected normal landing), SL (soft landing).

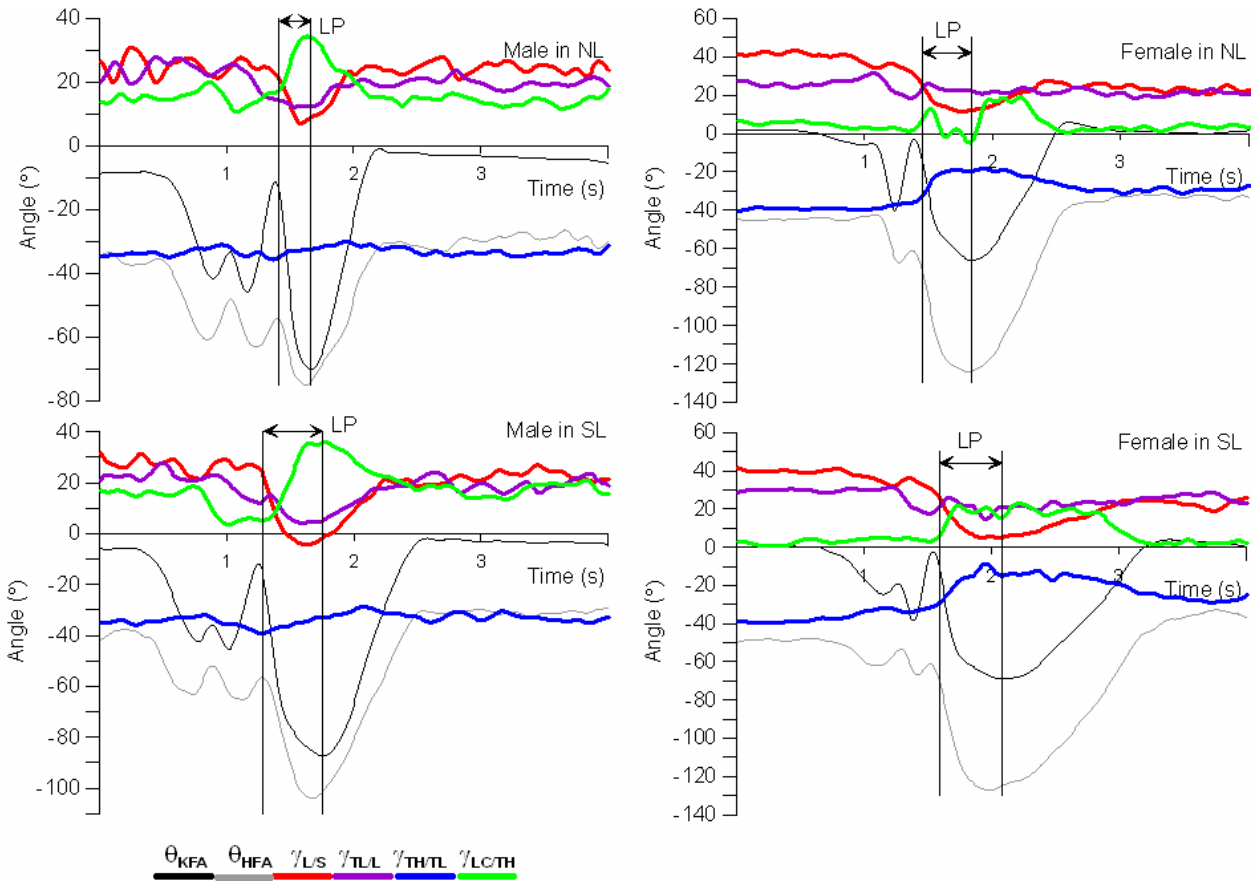

Figure 4-2. Representative kinematics of each spinal region ($\gamma_{L/S}$: lumbar regional angle relative to the sacral region, $\gamma_{TL/L}$: thoracolumbar regional angle relative to the lumbar region, $\gamma_{TH/TL}$: thoracic regional angle relative to the thoracolumbar region, $\gamma_{LC/TH}$: lower cervical regional angle relative to the thoracic region), knee flexion angle (θ_{KFA}) and hip flexion angles (θ_{HFA}) in one male and one female subjects during NL and SL conditions. For the direction of motion, negative slope represents the flexion motion. Flexion of a spinal region means the forward rotation of a region relative to the adjacent lower region. Note: LP (landing phase), NL (self selected normal landing), SL (soft landing).

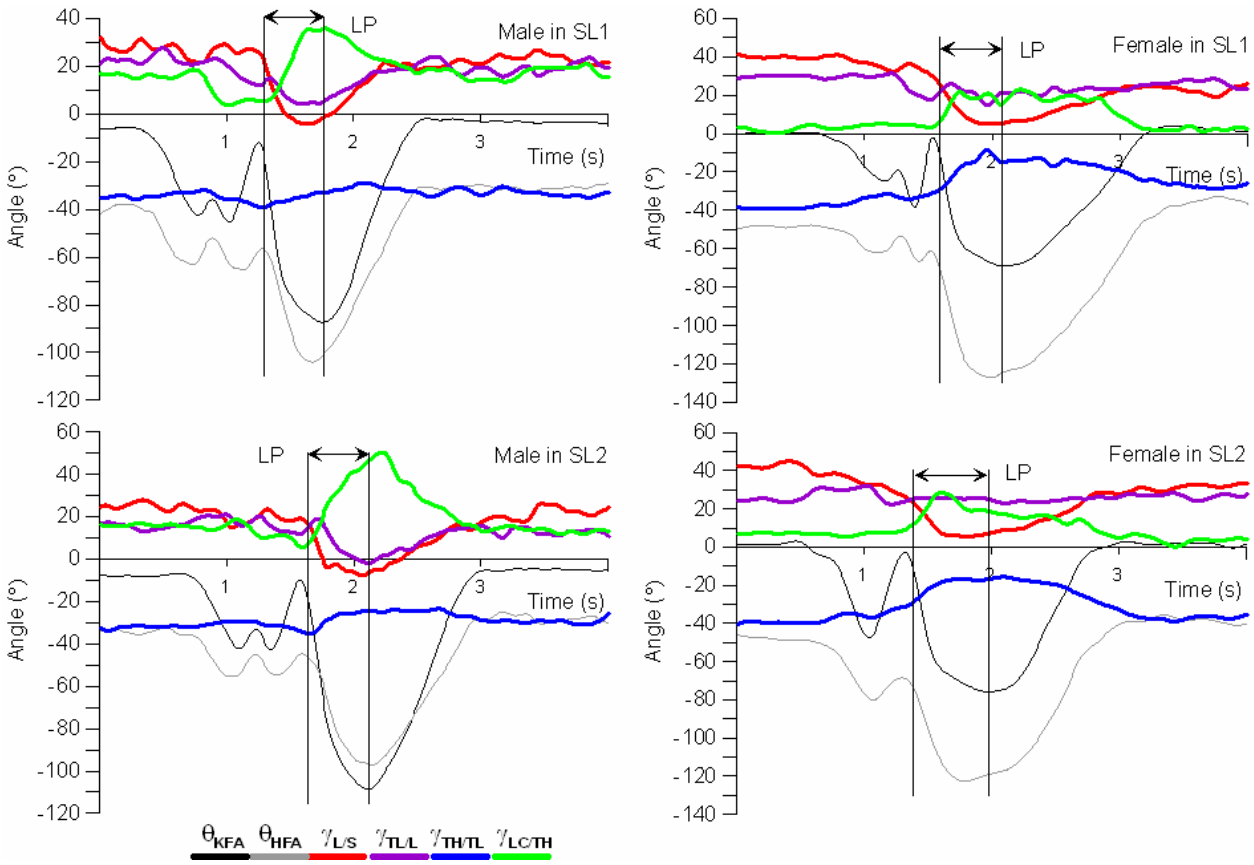

Figure 4-3. Representative kinematics of each spinal region ($\gamma_{L/S}$: lumbar regional angle relative to the sacral region, $\gamma_{TL/L}$: thoracolumbar regional angle relative to the lumbar region, $\gamma_{TH/TL}$: thoracic regional angle relative to the thoracolumbar region, $\gamma_{LC/TH}$: lower cervical regional angle relative to the thoracic region), knee flexion angle (θ_{KFA}) and hip flexion angles (θ_{HFA}) in one male and one female subjects during SL1 and SL2 conditions. For the direction of motion, negative slope represents the flexion motion. Flexion of a spinal region means the forward rotation of a region relative to the adjacent lower region. Note: LP (landing phase), SL1 (soft landing before the fatigue procedures), SL2 (soft landing after the fatigue procedures).

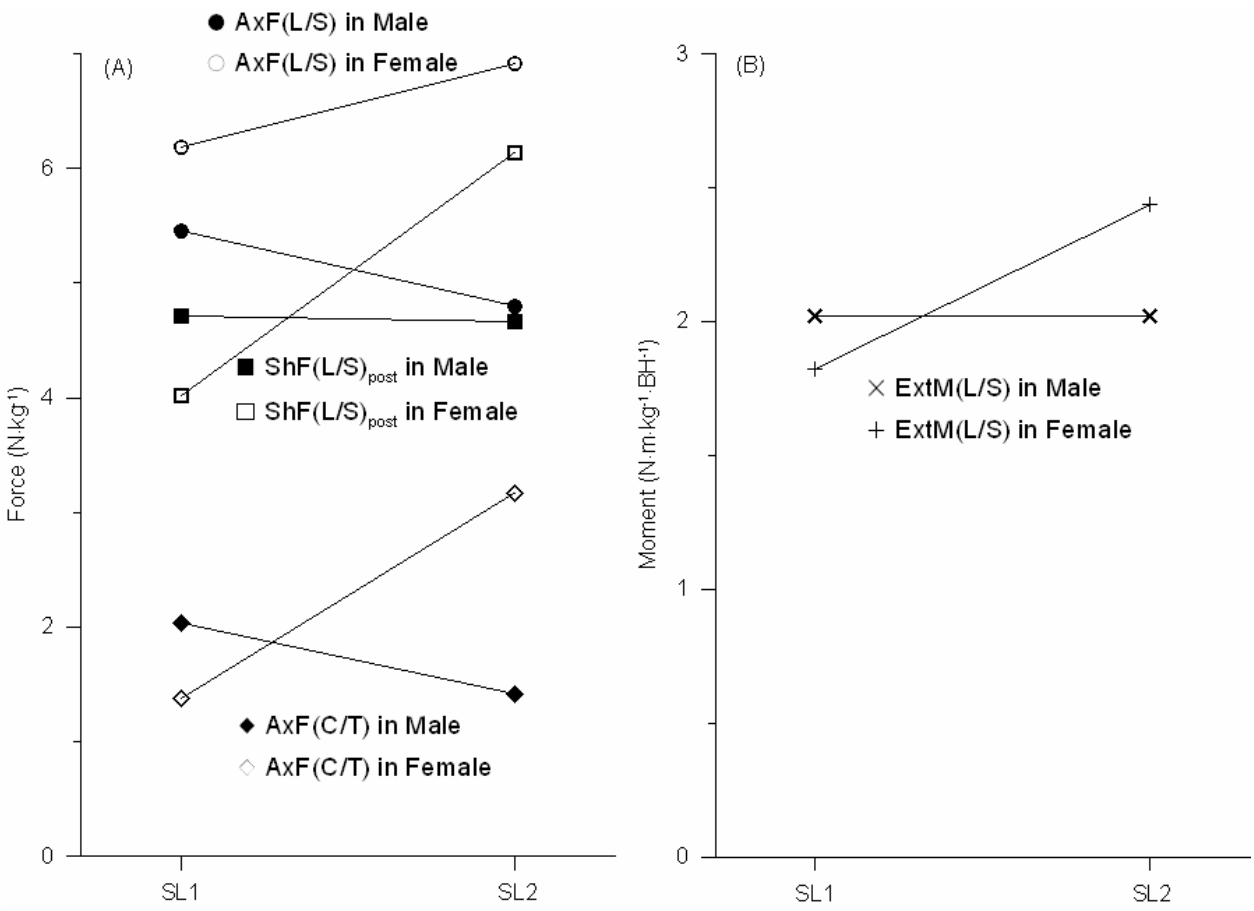

Figure 4-4. Significant interactions of the kinetic variables at L/S and C/T junctions between gender and fatigue level: (A) AxF(L/S) was increased from SL1 to SL2 condition in females, but not increased in males; $ShF(L/S)_{post}$ was increased from SL1 to SL2 condition in females, but not increased in males; AxF(C/T) was increased from SL1 to SL2 in females, but not increased in males, (B) ExM(L/S) was increased from SL1 to SL2 condition in females, but not increased in males. Note: SL1 (soft landing before the fatigue procedures), SL2 (soft landing after the fatigue procedures).

CHAPTER 5
DISCUSSION

Effects of Landing Technique

Previous studies on landing mechanics focused mostly on biomechanical characteristics of lower extremities. Majority of these studies focused on defining injury mechanism (Fagenbaum & Darling, 2003; Wikstrom et al., 2006), gender differences (Kernozek et al., 2005) and conditional variability (Schot et al., 2002) associated with knee and ankle joints. To our best knowledge, no attempt has been made to evaluate the mechanical characteristics of upper body segments above the hip joints during drop landings.

The landing variables used in the current study identified the overall landing characteristics including knee and hip joints. The landing technique used in soft landing significantly decreased the PVGRF, extended the landing phase, exhibited flexed landing postures of knee and hip joints, and exhibited more flexion motion of knee and hip joints. Additionally, female subjects demonstrated a greater degree of hip flexion than male subjects, and this may indicate that females demonstrate absorbing type landing without any different instructions.

Horita et al. (2002) proposed two types of landing performances with regard to the pre-landing motion of the knee joint. The proper pre-landing movement was characterized by knee flexion just before touchdown, which is associated with a high initial joint stiffness coupled with high joint power. This was called *bouncing type* in their study, which is similar to plyometrics. On the other hand, an inadequate pre-landing movement, associated with incomplete knee flexion induced subsequent deep-knee flexion after touchdown, was called the *absorbing type* of landing and was regarded as a poor strategy. The absorbing movement comes too late and demands longer contact time and lower takeoff velocity. A recent study by Park et al. (2006a) revealed that the soft landing strategy used by males was close to the bouncing type which is fit

for the effective subsequent movements, and that of females was close to absorbing type of landing which is just pursuing the dissipation of increased loads from ground impact.

The lumbar region was in a flexion posture immediately before touchdown and reached its maximal flexion around the end of landing phase, and this was followed by a rapid extension to neutral after the landing phase. For the purpose of discussion, the neutral angle of a spinal region is defined as the spinal regional angle recorded during a standing posture at the end of a landing trial. Flexion motion of a spinal region refers to forward rotation of the spinal region relative to the lower adjacent spinal region. The thoracic region started an extension motion after touchdown followed by a flexion to neutral toward the end of landing phase. However, the thoracic spine moved more gradually compared to the movement of lumbar region. The thoracolumbar region had a short duration of slight extension around touchdown and reached its initial peak at the beginning of the landing phase followed by a flexion motion similar to the lumbar region. The lower cervical region exhibited an extension motion around touchdown, and reached its full extension during the landing phase followed by a rapid flexion to neutral. Most spinal regions demonstrated biphasic motions (i.e., extension followed by flexion or flexion followed by extension), but the thoracolumbar region demonstrated multiphasic motion during the landing phase of drop landing.

The biphasic pattern of spinal motion has been observed during walking. Crosbie et al. (1997) studied the patterns of spinal motion during walking using a model including upper and lower trunks, lumbar and pelvis segments. They used a surface marker in each spinal segment on the back surface of the subjects. The pattern of flexion/extension of each segment was generally biphasic throughout the gait cycle. The pelvis rotated into negative pelvic tilt at heel strike, and this was followed by a counter-motion to a maximum positive pelvic tilt in the single support

phase. The lumbar spine showed maximum flexion at heel strike, and this was followed by a rapid extension to neutral until the single support phase. The lower thoracic segment extended maximally at heel strike, and returned to a neutral at mid-stance, then extended through the late stance phase.

Kinematics of the spinal column after touchdown of drop landing was addressed with touchdown angle and extension ROM variables. Relative to research question Q1 (any gender differences in spinal kinematics?), the null hypothesis 1a was rejected by the finding of gender differences in thoracolumbar touchdown angle and extension ROM of the thoracic region.

Females demonstrated a more extended posture in the thoracolumbar region at landing and more extension motion in the thoracic region during the landing phase compared to males. The greater thoracic region extension may be related to the greater hip flexion in females (balance control). An extended lower extremity landing posture and more flexion motion of lower extremity joints are considered as specific landing characteristics in females (Decker et al., 2003). According to the work-energy relationship, the average vertical GRF experienced by the subject during a drop landing depends on the vertical displacement of whole body CG (center of gravity) during the landing phase. Because an increased flexion motion of the hip joint will lower the body CG, the action of the thoracic region may compensate the lowered CG during the landing phase. As a result, the energy absorbing procedure in the landing phase may become less effective in females because of the movements of spinal column against the energy absorption such as the extension motion of the thoracic region.

Relative to research question Q2 (any changes in spinal kinematics by the different landing technique?), the null hypothesis 2a was rejected by the finding of significant changes in $\gamma_{P(TH/TL)}$, $\gamma_{P(LC/TH)}$, and $\gamma_{TD(TH/TL)}$ across different landing techniques. When going from NL to SL

condition, the thoracic and lower cervical regions exhibited more extension motion. Males exhibited more flexed landing posture in the thoracic region from NL to SL, while females did not or showed extended posture.

Active movement of each spinal region in both landing conditions may indicate that the spinal column is actively involved in energy absorption during the landing phase. Without analyzing the flexion ROM in each spinal region (only extension ROMs were analyzed), the flexion ROM of each spinal region could not be apparently differentiated with each other. However, the very small initial peaks of extension observed in the lumbar region across different landing techniques indicate that the lumbar region exhibited flexion motion in most cases. Additionally, during a soft landing, the thoracic and lower cervical regions demonstrated active extension motions against the lumbar motion. This may suggest that the entire spinal column is involved in energy absorbing procedure during the NL condition, while only part of spinal column is involved during the SL condition. The partial involvement is more apparent in females during the soft landing. Futhermore, the gender differences in thoracic landing posture during soft landings may suggest that the spinal column is less involved in energy absorbing procedure of females during soft landings (see Figure 4-4).

The results also suggest that the thoracolumbar region could be highly stressed by the simultaneous motions of thoracic and lower cervical extension and lumbar flexion during the landing phase of soft landing and is more pronounced in females. Schache et al. (2003) performed a kinematic study of the lumbo-pelvic-hip complex during running to define the gender differences. They found that females displayed a shorter stance time, swing time, stride time and stride length, and a higher stride rate than males. Mean waveforms were different in the peak-to-peak oscillations and the offset of pelvis anterior/posterior tilt. Females displayed greater

amplitudes of lumbar spine lateral bend and axial rotation, pelvis anterior/posterior tilt, obliquity and axial rotation, and hip adduction/abduction than their male counterparts. The mean positions of anterior pelvic tilt across the running cycle were 20.2° for females and 16.9° for males. The prevalence of pelvic-femoral stress fractures in female runners might be explained by these findings (Bennell et al., 1996; Pavlov et al., 1982). The greater amplitude of lumbar spinal movements in females during drop landings is similar to the findings observed in running.

The increased extension movement of the thoracic region relative to the thoracolumbar region from NL to SL condition suggests that the soft landing procedure intended to decrease lower extremity loads may be a risk factor for developing thoracolumbar or upper lumbar degeneration and spondylolysis in physically active individuals. Degenerative disc disease and spondylolysis are the most common structural abnormalities associated with low-back pain in athletes. Disc degeneration appears to be influenced by the type and intensity of the sport. Videman et al. (1995) demonstrated that weight lifters have a higher rate of and more severe degenerative changes in the upper lumbar spine, whereas back problems in soccer players are almost exclusively in the L4 to S1 levels. Cappozzo et al. (1985) found that, when a person performed half-squat exercises with weights approximately 1.6 times body weight, compressive loads across the L3/L4 motion segment were about 10 times body weight. The prevalence of spondylolysis in athletes is variable, but some sports appear to be associated with a higher prevalence rate. Rossi and Dragoni (1990) reported a rate of 43% in divers, 30% in wrestlers, and 23% in weight lifters. Although the exact mechanism for the development of spondylolysis is not known, there is some suggestion that it may be a fatigue fracture following repeated hyperextension.

Joint resultants at L/S and C/T junctions after touchdown of drop landing were addressed with the L/S and C/T kinetic variables. Relative to research question Q1 (any gender differences in L/S or C/T kinetics?), the null hypothesis 1b was rejected by the finding of significant gender differences. Posteriorly directed shear force of the trunk segment at C/T junction was significantly greater in male subjects than in female subjects during NL, but no gender difference for the SL condition was found. Females demonstrated decreased $ShF(C/T)_{ant}$ from NL to SL condition, while males did not demonstrate any changes in $ShF(C/T)_{ant}$ across landing techniques (see Figure 4-1). $ShF(C/T)_{ant}$ of females was greater than that of males during NL, but no difference between genders was found for the SL condition. Increased posteriorly directed shear force in males and increased anteriorly directed shear force in females for the NL condition can place much stress on supporting anatomical structures.

The cervical spine can be injured due to increased shear forces at the lower cervical region by a whiplash. During a whiplash, hyperextension of head and neck is the basic mechanism for cervical spine injury and it commonly occurs in the rear-end impact in motor vehicle accidents (Luan et al., 2000). Studies of the natural history of whiplash-associated disorders have suggested that chronic pain with continued symptoms develops in 6-33% of acutely injured patients (Hildingsson & Toolanen, 1990). Previous biomechanical studies have focused on injury mechanisms of the cervical facet joints and the intervertebral discs as potential structures to develop whiplash-associated disorders. Following a whiplash, the lower cervical spine experiences complex loading consisting of an extension moment, posterior shearing and compressive forces. This loading pattern has been hypothesized to injure the cervical intervertebral disc and facet joints (Panjabi et al., 2004).

Facet joints of the lower cervical spine can be compressed by the increased posterior shear force, and a capsular ligament strain can be distracted by the increased anterior shear force during the NL condition. Pearson et al. (2004) evaluated peak facet joint compression and capsular ligament strain using a whole cervical spine specimen with muscle force replication and a bench-top trauma sled to simulate whiplash of increasing severity. Peak facet joint compression was greatest at C4/C5 and reached over physiologic limits during the 3.5g simulation (low-level acceleration). Capsular ligament strains exceeded the physiologic strains at 6.5g and were largest at C6/C7 during the 8g simulation (high-level acceleration). They concluded that peak facet joint compression occurred at maximum intervertebral extension, whereas peak capsular ligament strain was reached as the facet joint was returning to its neutral position after the maximum intervertebral extension. The greater degree of posteriorly directed shear force in males than in females may indicate that male subjects could be in risk of increased facet joint compression, while females could be in risk of increased capsular ligament strain due to the increased anteriorly directed shear force during NL condition.

Panjabi and co-workers (2004) have studied a whole cervical spine model on a bench-top sled with muscle force replication and a surrogate head to simulate rear-end impact. They underwent standard flexibility testing to determine the sagittal disc deformation with the various acceleration of the bench. They observed the greatest strain in the posterior 150° fibers running posterosuperiorly at C5/C6. They also observed increased disc shear strain at the posterior region and increased axial deformation at the anterior region of the disc at C5/C6. They concluded that the cervical intervertebral disc is at risk for injury during a whiplash because of excessive 150° fiber strain, disc shear strain, and anterior axial deformation. The injury of the cervical spine by

facet joint distraction may occur at the level lower than the cervical injury by the facet compression.

Relative to research question Q2 (any changes in L/S or C/T kinetics by the landing technique?), the null hypothesis 2b was rejected by the finding of significant changes in all kinetic variables except for $ShF(C/T)_{ant}$ across different landing techniques. During the SL, all kinetic variables except for $ShF(C/T)_{ant}$ decreased significantly comparing to NL condition. This may indicate that the soft landing technique used in the current study is an effective way to decrease the overall loads applied to L/S and C/T junctions. However, $ShF(C/T)_{ant}$ was not decreased by the soft landing.

The greatest loads at L/S junction were AxF(L/S) and $ShF(L/S)_{post}$, while the $ShF(C/T)_{post}$ was the greatest load at C/T junction during drop landing. Callaghan et al. (1999) conducted a biomechanical study using two models to estimate loads applied to the L4/L5 level during walking task: linked segment model with EMG technique and rigid segment model with inverse dynamic technique. The joint loading at L4/L5 calculated by the EMG model resulted in large increases in the maximum compressive forces (3.5 times of body weight), compared with the joint reaction forces calculated using inverse dynamics (1.0 times of body weight). Including the muscular component resulted in a more than three-fold increase in joint load. However, the joint shear forces (anterior/posterior, lateral) obtained using the two techniques were quite similar. The peak compressive axial forces at L/S junction were 8.5 and 5.8 times of body mass for NL and SL conditions, respectively. The posteriorly directed shear force at L/S junction from the current study (9.7 × body mass in NL, 4.5 × body mass in SL) was much higher than the values for their walking trials (EMG model: 0.18 × BW, Inverse dynamics model: 0.19 × BW). In the study of Callahan et al. (1999), there was a peak flexor moment at heel contact followed by a

peak extensor moment around toe-off. During a faster speed gait, the flexion/extension moment at L4/L5 shifted to the extension side and demonstrated a high extensor moment around toe-off. Callahan et al. concluded that the loads and motions for the lumbar spine during gait depended on the walking speed. Increasing walking speed increases the lumbar spine ROM, activation of spinal and trunk muscles, and anterior/posterior shear forces. Likewise, the loads and motions for the lumbar spine during drop landing depend on the landing technique which controls the involvement of body segments including lower extremity joints and spinal regions.

One of the limitations of this study is that sagittal spinal kinematics relative to the adjacent spinal region based on the spinal skin markers cannot precisely determine the vertebral motions in each spinal region. Additionally, there might be some errors in kinetic measures, because the locations of joint center and spinal junction (used for joint resultants computations) were estimated using surface markers instead of determined using radiographic imaging techniques. Mechanical characteristics of the lower extremities were assumed to be symmetrical and only the GRF data collected from the left leg were used to determine joint resultants at the L/S and C/T junctions. Despite these limitations, the results from this study provide insight into the spinal movement during two different landing techniques.

Effects of Knee Joint Muscles Fatigue

The purpose of the second part of this study was to determine whether spinal mechanics were affected by lower extremity fatigue during drop landings. The overall landing mechanics did not change significantly by the presence of knee joint muscles fatigue in the current study. Instead, a significant gender difference in landing posture was found in the knee joint. Females landed with a more extended knee joint posture than males which is consistent with the findings reported in a previous study (Decker et al., 2003). These authors found that females demonstrated a more erect landing posture and utilized greater hip and ankle joint range of

motions and maximum joint angular velocities when compared to males. Females exhibited greater energy absorption and peak powers from the knee extensors and ankle plantar-flexors compared to males. Energy absorption contributions revealed that the knee extensor was the primary shock absorber for both genders. The ankle plantar flexor was the second largest contributor to energy absorption for the females and the hip extensor was for the males. The different shock absorption strategy used in females was proposed to provide a greater potential risk for non-contact ACL injury for females under certain landing conditions.

Kinematics of the spinal column after touchdown of drop landing was addressed with touchdown angle and extension ROM variables. Relative to research question Q3 (any gender differences in spinal kinematics?), the null hypothesis 3a was rejected by the finding of significant gender differences in thoracolumbar touchdown angle and extension ROM of the thoracic region. Females demonstrated a more extended posture in the thoracolumbar region at landing and more extension motion in the thoracic region during the landing phase compared to males. This is quite similar to the findings of lower extremity joints in the study of Decker et al (2003).

The greater extension motion of the thoracic region is supposed to compensate the extended postures in the knee joint and thoracolumbar region at landing in females (balance control). An extended lower extremity landing posture and more flexion motion of lower extremity joints are considered as specific landing characteristics in females (Decker et al., 2003). The extended postures of the knee and thoracolumbar region will raise the CG location of the whole body and increase the stiffness of these regions compared to their male counterparts. The increased extension motion of the thoracic region will further raise the body CG and may

contribute to an increase in stiffness of the whole body during the landing phase for females during the fatigue landing condition.

Kujala et al. (1997) studied the impact of lumbar flexibility on low-back pain in a 3-year longitudinal study. They examined lumbar ROM in a group of adolescent athletes and nonathlete controls across both genders. Neither groups had previous low-back pain, nor were lumbar measurements performed during episodes of pain. While no differences were detected between athletes and nonathlete controls in males, female nonathletes exhibited greater overall lumbar ROM and lower lumbar ROM than did female athletes. They suggested that decreased ROM in the lower lumbar segments and decreased maximal lumbar extension as the predictive factors of low-back pain in women, because the girls within the lowest quartile of maximal lumbar extension developed 3.4 times the chance of having pain lasting more than one week. Sward et al. (1990) studied lumbar mobility, in relation to back pain, in male athletes, but no correlation was found between spinal flexibility and back pain. Both previous studies support that high mobility of the lumbar region is essential to maintain the lower-spinal integrity in females, and the active women or female athletes who were involved in vigorous activities demanding much spinal motion will have more chance to develop limitation of spinal movement and low-back pain. Gender differences in kinematic characteristics observed in the current study suggest that higher loads due to the hyperextension of the thoracic region could be placed on the thoracolumbar region of females during soft landing and higher incidence of low-back pain may be found in female athletes or active women who perform repeated jumps and landings. Without enough flexibility and strength of thoracolumbar to lumbar region, females are likely to be at risk to develop spinal injury or back pain due to repeated soft landings.

The kinematic pattern of each spinal region in the current study is comparable to the findings observed in previous walking and running studies. Crosbie et al. (1997) evaluated spinal kinematics during walking tasks. The lumbar spine showed maximum flexion at heel strike followed by a rapid extension to neutral until the single support phase. The lower thoracic segment extended maximally at heel strike, and returned to a neutral posture at mid-stance, then extended through the late stance phase. Schache et al. (2002) observed the kinematics of pelvis and lumbar spine during running trials. The lumbar spine flexed slightly and the pelvis posteriorly tilted slightly during the loading phase. The lumbar spine was extended while the pelvis was anteriorly tilted during the swing phase. Both lumbar spine and pelvis displayed a biphasic movement pattern that corresponds to one phase per each step. The biphasic patterns of pelvis, lumbar and thoracic spines in both previous studies resemble the movement pattern of each spinal region during soft landing trials.

In the current study, the lumbar region started a flexion motion before the touchdown and reached its peak around the end of landing phase, and this was followed by a rapid extension to neutral after the landing phase. For the purpose of discussion, the neutral angle of each spinal region in the current study is defined as the spinal regional angle recorded during the standing posture at the end of a land trial. The thoracic region started an extension motion around touchdown followed by a gradual flexion to neutral toward the end of landing phase. The thoracolumbar region had a short duration of slight extension immediately after touchdown and reached its initial peak extension at the beginning of the landing phase followed by a flexion motion similar to the lumbar region. The lower cervical region started an extension motion around touchdown and reached its peak extension around the end of the landing phase followed by a rapid flexion to neutral. Generally, most spinal regions demonstrated biphasic motions

during the landing phase of soft landing, but the thoracolumbar region exhibited multiphasic motion. The kinematic pattern of spinal column observed in the soft landing trials suggests that the thoracolumbar region could be highly stressed by simultaneous lumbar flexion and thoracic extension during a greatly loaded landing phase.

Relative to research question Q4 (any changes in spinal kinematics during soft landing by the fatigue procedures?), the null hypothesis 4a was supported by a lack of significant kinematic differences in any spinal regions between SL1 and SL2. Fatigue procedures applied to knee flexors and extensors in the current study was introduced by repeated bouts of maximal work of target muscles, because muscular fatigue has been defined as the development of less than the expected amount of force as a consequence of continuous voluntary muscle contractions (McCully et al., 2002). The expression of muscle fatigue as a percent reduction in torque output has been used as a fatigue index (Katsiaras et al., 2005). The fatigue index is typically computed as the ratio of the average torque of the last five repetitions to the average torque of the first five repetitions in a 30-repetition maximum effort trial. However, this quantification of muscle fatigue has been questioned in terms of its reliability. Because torque variability within the initial five contractions has a potentiation effect, muscle fatigue appeared to be underestimated when the first five repetitions were factored into the fatigue index formula (Neptune et al., 1997). Since torque output during maximal contractions at 3.14 $rad \cdot s^{-1}$ has been shown to increase from repetitions four to as high as 10 (Wretling & Henriksson-Larsen, 1998), a more accurate estimate of muscle fatigue can be obtained with the highest consecutive five repetitions as the subject's best performance (Pincivero et al., 2003).

Muscular fatigue is manifested as a progressive decline in power output, and the magnitude of which is largely determined by the duration of the intervening recovery periods

(Glaister, 2005). Thirty minutes of running was included during the fatigue procedures of the current study for the purpose of delaying the recovery period. In general, hypoxic condition is associated with the increased rates of fatigue. For example, under the hypoxic condition during the long-distance running, the decreased ability to perform cycled maximal sprints was associated with an increased accumulation of blood lactate, a reduced oxygen uptake, and an increased rate of muscular fatigue. Oxygen availability mediates its effect on multiple sprint performance by influencing the magnitude of the aerobic contribution to ATP (adenosin triphosphate) resynthesis during work periods, and the rate of phosphocreatinine resynthesis during intervening rest periods (Balsom et al., 1994).

Data collected from 3 female and 1 male subjects were not included in the statistical analyses because their fatigue indices did not increase due to the fatigue procedure. It is possible that the intensity of fatigue procedure was not high enough for these 4 individuals. More exclusion of female subjects than male subjects from the current fatigue procedure may be explained by the fact that females are less susceptible to muscular fatigue. Hunter and Enoka (2001) observed that females sustained the maximal voluntary contraction for a much longer period of time (118% longer) than males and during low-level contraction (20% of the maximal voluntary contraction). It has been suggested that muscle fibers in females possess a relatively greater oxidative capacity than males, which would enhance the respiratory capacity of the contracting muscles. On the other hand, males may possess an inherently greater ability to generate more force than females, which may be related to a significantly greater proportion of fast-twitch muscle fibers in the skeletal muscle (Fulco et al., 2001).

No significant changes in landing posture in lower extremity joints due to the fatigue procedures may indicate that stretch reflex activity of the knee joint muscles was not affected by

the fatigue procedures and the fatigue level was not high enough to cause any kinematic changes or to be detected by the current measurements. The ground contact phases of running, jumping and hopping are examples of the stretch-shortening cycle (SSC) type of exercises for leg extensor muscles. Duing a SSC, the preactivated muscle is first stretched (eccentric action) and then followed by the shortening (concentric) action. The stretched phase is mediated by the reflex activity of the extensor muscles before the ground impact of the foot during jumps and landings (Horita et al., 2002). As the SSC demands a strong mechanical loads to the skeletal muscles, its influence on the reflex activation is essential to perform the fast and smooth SSC.

Intensive SSC-type exercise results in reversible muscle damage. This is associated with delayed-onset muscle soreness, and with proprioceptive and neuromuscular impairments that may last for several days. These neuromuscular perturbations are typically associated with changes in muscle mechanics and activation that result in major consequences on joint and muscle stiffness regulation in SSC-type performance. This performance deterioration is called as *neuromuscular fatigue* and subsequent long-term recovery will take place in a bimodal fashion. In this bimodality, the acutely induced reduction in electromyographic activity (maximal voluntary contraction) is followed by a short-term recovery (within 2 hours), which is in turn followed by a secondary reduction with longer lasting recovery (1-2 days post-exercise) (Nicol et al., 2006).

Joint resultants at L/S and C/T junctions after touchdown of drop landing were addressed with the L/S and C/T kinetic variables. Relative to research question Q3 (any gender differences in L/S or C/T kinetics?), the null hypothesis 3b was rejected by the finding of gender differences in AxF(L/S), $ShF(L/S)_{post}$, ExtM(L/S) and AxF(C/T). Males exhibited greater $ShF(L/S)_{post}$,

ExtM(L/S), and AxF(C/T) than females during SL1, but females showed greater $ShF(L/S)_{post}$, ExtM(L/S), and AxF(C/T) than males during SL2 condition (see Figure 4-4).

Significant kinematic gender differences found in the current study are extended landing posture of the knee joint and thoracolumbar junction and more extension motion of thoracic region during the landing phase in females. These findings may suggest that the knee joints and lumbar to thoracolumbar region are stiffer during the landing phase in females when comparing to males. This also suggests that males utilize the spinal column in energy absorption in addition to lower extremities. However, despite the low fatigue level of the Knee joint muscles, females exhibited significant increases in joint resultants at L/S and C/T junctions. Because females rely more heavily on lower extremity action in energy absorption during the landing phase, their landing mechanics are likely to be more affected by lower extremity fatigue than their male counterparts. Just like the interaction between leg stiffness and reflex activities (Horita et al., 1996), reflex activity of lumbar (to thoracolumbar region) spinal extensors also may play a role to maintain lumbar stiffness during the initial landing phase.

Relative to research question Q4 (any changes in L/S or C/T kinetics during soft landing by the fatigue procedures?), the null hypothesis 4b was rejected by the finding of significant changes in AxF(L/S), $ShF(L/S)_{post}$, ExtM(L/S) and AxF(C/T) from SL1 to SL2 condition in females. Females exhibited increased AxF(L/S), $ShF(L/S)_{post}$, ExtM(L/S) and AxF(C/T) when going from SL1 to SL2 condition, while males did not (Figure 4-4 on p. 91).

The significant increases in joint resultants at L/S and C/T junctions due to the fatigue procedure in females illustrate the significant role of spinal column in energy absorbing procedure during drop landing. However, the spinal column may play a different role in energy absorbing procedure in each gender. A recent study by Park et al. (2006b) also suggested the

important role of lumbar region during the landing phase. They studied the effects of limited lower back motion on soft landing mechanics of lower extremity joints with subjects wearing soft and hard low-back braces. Limited spinal motions by the brace caused alterations in knee and hip joint motions during the landing phase and an increase in impact force. Females demonstrated increased knee extensor moment and males showed increased axial force at the hip joint in the hard brace condition.

Increased axial force at C/T junction in females during the landing with knee joint muscles fatigue can place much stress on the cervical spine in some instances. Axial loading has been reported as a mechanism of catastrophic cervical spine injuries in football players (Torg et al., 1990). Nightingale et al. (1996) reported that straightening of the cervical spine before injury may be another necessary element of the compressive injury mechanism. Cervical spine trauma accounts for about 25% of entire spine injuries in sports (Leidholt, 1963). Cervical spine injury is more common in some sports than in others: football, gymnastics, rugby, baseball, lacrosse, judo, skiing, jumping on trampolines, and diving (Torg et al., 2002). Traditionally, hyperflexion and hyperextension have been implicated as the primary mechanisms of cervical spine injuries (Gehweiler et al., 1979; Macnab, 1964; Paley & Gillespie, 1986).

Torg (1987) provided the injury mechanism to the cervical spine in football player. A typical situation is that of a defensive back making a tackle involving contact with the other player with the top of the helmet. The injury mechanism involves axial loading with an element of buckling. With the neck in a neutral position, the cervical spine is extended as a result of the normal cervical lordosis. When the neck is flexed to 30°, the cervical spine becomes straight. When a force is applied to the vertex, the energy inputs are transmitted along the longitudinal axis of the cervical spine. The cervical spine loses its ability to dissipate force and being

compressed between the abruptly decelerated head and the force of the upcoming trunk. This is a typical example of cervical spine injury with the axial loading as the injury mechanism, which is characterized as a clustering of injuries in the middle part (third and fourth) of the cervical spine. Although the axial force at C/T junction during drop landings is much lower than the direct compression of vertex during contact sports, repeated landings with fatigued knee joint muscles may increase the chance of injuries to the cervical spine in females.

Luan et al. (2000) analyzed neck kinematics and loading patterns with high-speed X-ray video cameras during rear-end impact simulation study using a cadaver body. They observed compression, tension, shear, flexion and extension at different cervical levels during different stages of the whiplash event. They reported that compression of the neck is due to the straightening of the thoracic spine and possibly the upward ramping of the torso during the initial stage of the impact. Likewise, straightening of either cervical spine or thoracic spine will increase the axial compressive force more at the lower cervical region during the soft landing procedures. Accordingly, to avoid cervical injuries, individuals with spinal degeneration or surgical fusion in the cervical spine may need to refrain from repeated landing type activities.

In addition to the limitations mentioned previously, another limitation of this study is that the effects of fatigued muscle could not be precisely differentiated between flexor and extensor of the knee joint.

CHAPTER 6
CONCLUSION

Effects of Landing Technique

The spinal column is actively involved in energy absorbing procedure during the landing phase. However, males may use the the spinal column as the active weight bearing segment during the landing phase with less motion comparing to females. Females may demand spinal column in their energy absorbing procedure in landing phase but they need more extension motion comparing to males. Vigorous extension motions of the thoracic and lower cervical regions during soft landing suggest that the spinal column is less involved in energy absorption, while the entire spinal column is involved during the normal landing condition. The thoracolumbar region is likely to be highly stressed by the simultaneous motions of thoracic extension and lumbar flexion in females during the landing phase of soft landing. The cervical spine can be injured by the great posterior shear force in males through the facet joint compression, and can be injured by the great anterior shear force in females through the capsular ligament distraction during drop landings.

Effects of Knee Joint Muscles Fatigue

Despite the low level of the knee joint muscles fatigue, females demonstrated increased joint resultants at L/S and C/T juctions. Because extended landing postures of knee joint and thoracolumbar region were not compensated effectively by the increased motion of thoracic region during the landing phase of females, spinal mechanics are easily affected by the small changes in fatigue. It is also speculated that energy absorption in females could be mostly moderated by the extensor actitivities of knee joints. Additinally, the thoracolumbar region could be stressed when simultaneous lumbar flexion and thoracic extension occurred during a highly loaded initial landing phase of soft landings, and it was more prominent in females.

Conclusively, knee joint muscle fatigue during soft landing of females can be a risk factor to the lower lumbar and cervical spine injuries because of increased joint resultants.

CHAPTER 7
FUTURE WORK

The landing task can be an ideal situation to evaluate spinal mechanics for the young and active population. The measurements used in the current study could be applied to individuals in different athlete and patient populations. In addition to dynamic X-rays, the movement of each spinal region may provide valuable information on roles played by spinal regions in dynamic tasks. The data collected from patient groups may provide insight into the spinal function for different populations, and could be useful for decisions relative to conservartive and surgical treatments.

The inverse dynamic approach used in the current study can be compared with other modeling approaches. The inverse dynamic approach can only be used to determine joint resultants. Other modeling approaches (e.g. EMG-optimization model) have to be used in order to estimate contact forces at a joint.

APPENDIX A
INFORMED CONSENT

The informed consent approved by the Institutional Review Board of the University of Florida.

IRB# *116-2006*

Informed Consent to Participate in Research

University of Florida
Health Center
Institutional Review Board
APPROVED FOR USE
From 3/26/06 Through 3/25/07
CJR

You are being asked to take part in a research study. This form provides you with information about the study and informs you of how your privacy will be protected. The Principal Investigator (the person in charge of this research) or a representative of the Principal Investigator will also describe this study to you and answer all of your questions. Your participation is entirely voluntary. Before you decide whether or not to take part, read the information below and ask questions about anything you do not understand. If you choose not to participate in this study you will not be penalized or lose any benefits to which you would otherwise be entitled.

1. **Name of Participant ("Study Subject")**

   _____

2. **Title of Research Study**

   Effect of Fatigue on Drop Landing Mechanics

3. **Principal Investigator and Telephone Number(s)**

   Soo-An Park, M.S., M.D., Doctoral student, 151 FLG, Department of Applied Physiology & Kinesiology, PO Box 118206, Gainesville, FL 32611-8206, parkas@ufl.edu
   Office: (352) 392-9575 ext 1321, Cell phone: (352) 262-7420

4. **Source of Funding or Other Material Support**

   University of Florida

5. **What is the purpose of this research study?**

   The purpose of this study is to examine the effects of knee flexor/extensor muscles fatigue on movement characteristics during the landing of a drop jump.

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6. What will be done if you take part in this research study?

You will be asked to visit the Biomechanics Lab in Florida Gym to participate in the experiment. During the visit, you will be asked to complete the following:

- If you don't have any history of cardio-respiratory and musculoskeletal diseases of lower extremities and spine, you will be asked to complete a Physical Activity Readiness Questionnaire (PAR-Q) to verify your appropriateness to attend the experiment.
- If you meet the appropriate requirements in PAR-Q, then you will be accepted as a subject in this study.
- Your height, weight, and sizes of your legs, thighs, and feet will be measured.
- You will be asked to warm up by fast walking for 5 minutes on a treadmill and performing stretching exercises.
- You will perform the forward bending and several practice landings. After that, you will perform a series of jumps. Specifically, you will be asked to perform three drop landings from a height of 50 cm. Next, you will be asked to perform 3 drop landings with soft landing technique.
- You will move to a dynamometer and perform 20 repetitions of maximal isokinetic knee flexion/extension exercise.
- You will be asked to run on the motorized treadmill for 30 minutes with a speed of 5 – 6 mph.
- Isokinetic torque of knee flexors/extensors will be measured again to verify the fatigue level of knee joint muscles.
- Finally, you will perform 3 drop landings using the soft landing technique.

If you have any questions now or at any time during the study, you may contact the Principal Investigator listed in #3 of this form.

7. If you choose to participate in this study, how long will you be expected to participate in the research?

You are expected to attend one data collection session which will last for about 2 hours.

8. How many people are expected to participate in this research?

We expect 60 people to participate in this research study.
This study is being done only here in the Biomechanics Laboratory.

9. What are the possible discomforts and risks?

There will be no greater risk involved in participating in this study than in common physical activities. The anticipated risks associated with this study would be no more than those associated with playing a sport in which running, jumping, and landing are required such as soccer, basketball or volleyball.

There could be soreness in leg and thigh muscles after the testing session. But this risk will be minor and it will not extend more than 2 to 3 days post testing.

This study may include risks that are unknown at this time.

Participation in more than one research study or project may further increase the risks to you. Please inform the Principal Investigator (listed in #3 of this consent form) or the person reviewing this consent with you before enrolling in this or any other research study or project.

Throughout the study, the researchers will notify you of new information that may become available and might affect your decision to remain in the study.

If you wish to discuss the information above or any discomforts you may experience, you may ask questions now or call the Principal Investigator or contact person listed on the front page of this form.

10a. What are the possible benefits to you?

You will not personally benefit from participating in this study.

10b. What are the possible benefits to others?

To exercise scientists, the findings may provide insights into movement coordination and control during a landing task when knee joint muscle becomes fatigue or weakened. The results may have implications to sports training and performance.

11. If you choose to take part in this research study, will it cost you anything?

No.

12. Will you receive compensation for taking part in this research study?

No. You will not be compensated monetarily for your participation in the study.

13. What if you are injured because of the study?

If you experience an injury that is directly caused by this study, only professional consultative care that you receive at the University of Florida Health Science Center will be provided without charge. However, hospital expenses will have to be paid by you or your insurance provider. No other compensation is offered. Please contact the Principal Investigator listed in Item 3 of this form if you experience an injury or have any questions about any discomforts that you experience while participating in this study.

14. What other options or treatments are available if you do not want to be in this study? Credits/course requirements ARE OFFERED to student for being in the study?

You have been invited to participate in this research project because you are a student and young healthy person free from any musculoskeletal diseases or injuries. Your participation as a subject in this study allows you to receive an extra-credit of 2% point toward the final grade of a class offered in HHP. The course has provisions for alternatives to participating in this study that will not affect your grade or hinder completing course requirements. The principal investigator (Soo-An Park) can provide you with a copy of the written statement regarding alternatives. You can decline to participate in this study without it affecting your grade or class standing.

If you believe that your participation in this study or your decision to withdraw from or to not participate in this study has improperly affected your grades(s), you should discuss this with the dean of your college or you may contact the IRB office.

15a. Can you withdraw from this research study?

You are free to withdraw your consent and to stop participating in this research study at any time. If you do withdraw your consent, there will be no penalty, and you will not lose any benefits you are entitled to.

If you decide to withdraw your consent to participate in this research study for any reason, you should contact Soo-An Park at (352) 392-9575 ext 1321 or (352) 262-7420.

If you have any questions regarding your rights as a research subject, you may phone the Institutional Review Board (IRB) office at (352) 846-1494.

15b. If you withdraw, can information about you still be used and/or collected?

If you withdraw this study, all the information about you will be discarded and not be used with any purposes.

15c. Can the Principal Investigator withdraw you from this research study?

You may be withdrawn from the study without your consent for the following reasons:

- You do not qualify to be in the study because you do not meet the study requirements.
    - Any history of cardio-respiratory and musculoskeletal diseases of lower extremities and spine.
    - Any 'yes' in your completed Physical Activity Readiness Questionnaire (PAR-Q) to verify your appropriateness to attend the experiment.
- You need a medical treatment not allowed in this study.
- The investigator decides that continuing in the study would be harmful to you.

16. How will your privacy and the confidentiality of your research records be protected?

Information collected about you will be stored in locked filing cabinets or in computers with

security passwords. Only certain people have the legal right to review these research records, and they will protect the secrecy (confidentiality) of these records as much as the law allows. These people include the researchers for this study, certain University of Florida officials, the hospital or clinic (if any) involved in this research, and the Institutional Review Board (IRB; an IRB is a group of people who are responsible for looking after the rights and welfare of people taking part in research). Otherwise your research records will not be released without your permission unless required by law or a court order.

If the results of this research are published or presented at scientific meetings, your identity will not be disclosed.

17. How will the researcher(s) benefit from your being in this study?

In general, presenting research results helps the career of a scientist. Therefore, the Principal Investigator may benefit if the results of this study are presented at scientific meetings or in scientific journals.

18. Signatures

As a representative of this study, I have explained to the participant the purpose, the procedures, the possible benefits, and the risks of this research study; the alternatives to being in the study; and how privacy will be protected:

_____ _____

Signature of Person Obtaining Consent Date

You have been informed about this study's purpose, procedures, possible benefits, and risks; the alternatives to being in the study; and how your privacy will be protected. You have received a copy of this Form. You have been given the opportunity to ask questions before you sign, and you have been told that you can ask other questions at any time.

You voluntarily agree to participate in this study. By signing this form, you are not waiving any of your legal rights.

_____ _____

Signature of Person Consenting Date

APPENDIX B
PRELIMINARY STUDY

One healthy active male (age: 21 yrs, mass: 80.7 kg, height: 179.7 cm) and one female (21 yrs, 59.9 kg, 168.3 cm) were tested in this pilot study. They were free from any musculoskeletal diseases or injuries which could influence spinal and lower extremity joints mechanics during physical activities, and signed informed consent approved by the IRB of the University of Florida. The experimental setup, testing procedures and data reduction were the same as what was described in Chapter 2. The peak extensor moments at the L/S junction were used to determine sample size for the current study (Table B-1).

Table B-1. Kinetic characteristics of L/S junction data for sample size justification.

		NL	SL1	SL2
AxF(L/S)	Mean (SD)	10.47 (0.25)	6.85 (0.57)	7.17 (2.47)
($N \cdot kg^{-1}$)	Male	10.65	7.25	5.42
	Female	10.29	6.44	8.92
$ShF(L/S)_{ant}$	Mean (SD)	1.80 (0.88)	0.84 (0.35)	0.87 (0.53)
($N \cdot kg^{-1}$)	Male	2.42	0.59	0.49
	Female	1.18	1.08	1.24
$ShF(L/S)_{post}$	Mean (SD)	7.35 (1.24)	4.37 (1.04)	6.11 (0.33)
($N \cdot kg^{-1}$)	Male	8.23	3.63	6.34
	Female	6.47	5.10	5.88
FlxM(L/S)	Mean (SD)	1.74 (0.48)	1.14 (0.09)	1.20 (0.08)
($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	Male	1.40	1.20	1.14
	Female	2.08	1.07	1.26
ExtM(L/S)	Mean (SD)	3.05 (0.38)	2.27 (0.15)	2.59 (0.19)
($N \cdot m \cdot kg^{-1} \cdot BH^{-1}$)	Male	3.32	2.16	2.72
	Female	2.78	2.37	2.45

Note: NL (self-selected normal landing), SL1 (soft landing before fatigue procedure), SL1 (soft landing after fatigue procedure), AxF (peak axial compressive force), $ShF_{ant(or\ post)}$ (peak ant. or post. shear force), FlxM (peak flexor moment), ExtM (peak extensor moment), (L/S) (for lumbosacral junction).

APPENDIX C
MANOVA AND ANOVA TABLES

Effects of Landing Technique

Table C-1. MANOVA table for landing variables, touchdown angle variables, extension ROM variables, and L/S and C/T kinetic variables in the study of landing technique effects.

Effect	Roy's Largest Root	F	Hypothesis df	Error df	Sig. (p)	Observed Power (a)
Landing variables						
Gender*	2.824	8.942	6	19	<0.001	1.0
Landing*	8.148	25.803	6	19	<0.001	1.0
Landing × Gender	0.797	2.523	6	19	0.057	0.7
Touchdown angle variables						
Gender*	0.807	4.236	4	21	0.011	0.85
Landing	0.287	1.509	4	21	0.236	0.39
Landing × Gender*	0.659	3.459	4	21	0.025	0.77
Extension ROM variables						
Gender*	0.788	4.136	4	21	0.013	0.84
Landing*	1.657	8.700	4	21	<0.001	0.99
Landing × Gender	0.290	1.523	4	21	0.232	0.39
L/S Kinetic variables						
Gender	0.269	1.076	5	20	0.403	0.3
Landing*	3.801	15.206	5	20	<0.001	1.0
Landing × Gender	0.410	1.639	5	20	0.196	0.45
C/T Kinetic variables						
Gender*	0.773	3.094	5	20	0.031	0.76
Landing*	4.065	16.259	5	20	<0.001	1.0
Landing × Gender*	0.899	3.597	5	20	0.018	0.83

* Significant main effect or interaction ($p<0.05$)

Table C-2. Univariate tests for the different landing variables in the study of landing technique effects.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power(a)
PVGRF	Gender	0.383	1	0.383	0.025	0.877	0.05
	Error	374.297	24	15.596			
	Landing*	861.538	1	861.538	90.268	<0.001	1.0
	Landing × Gender	1.231	1	1.231	0.129	0.723	0.06
	Error (Landing)	229.062	24	9.544			
$t_{(LP)}$	Gender	0.038	1	0.038	2.231	0.148	0.3
	Error	0.412	24	0.017			
	Landing*	0.230	1	0.230	26.355	<0.001	1.0
	Landing × Gender	0.003	1	0.003	0.395	0.536	0.09
	Error (Landing)	0.209	24	0.009			
$\theta_{TD(KFA)}$	Gender	276.001	1	276.001	3.789	0.063	0.46
	Error	1748.330	24	72.847			
	Landing*	136.988	1	136.988	23.797	<0.001	1.0
	Landing × Gender	7.388	1	7.388	1.283	0.268	0.19
	Error (Landing)	138.155	24	5.756			
$\theta_{P(KFA)}$	Gender	5.493	1	5.493	0.021	0.887	0.05
	Error	6361.497	24	265.062			
	Landing*	3711.240	1	3711.240	71.329	<0.001	1.0
	Landing × Gender	127.109	1	127.109	2.443	0.131	0.32
	Error (Landing)	1248.715	24	52.030			
$\theta_{TD(HFA)}$	Gender	10.351	1	10.351	0.035	0.853	0.05
	Error	7063.826	24	294.326			
	Landing*	348.407	1	348.407	24.873	<0.001	1.0
	Landing × Gender	1.489	1	1.489	0.106	0.747	0.06
	Error (Landing)	336.174	24	14.007			
$\theta_{P(HFA)}$	Gender*	2089.157	1	2089.157	15.422	0.001	0.96
	Error	3251.093	24	135.462			
	Landing*	4194.019	1	4194.019	50.000	<0.001	1.0
	Landing × Gender	227.643	1	227.643	2.714	0.113	0.35
	Error (Landing)	2013.138	24	83.881			

* Significant main effect or interaction ($p<0.05$)

Table C-3. Univariate tests for the touchdown angle variables in the study of landing technique effects.

Measure	Source	Type III SS	df	Mean Square	F	Sig. (p)	Observed Power (a)
Touchdown angle variables							
$\gamma_{TD(L/S)}$	Gender	241.231	1	241.231	1.625	0.215	0.23
	Error	3562.197	24	148.425			
	Landing	0.249	1	0.249	0.037	0.850	0.05
	Landing × Gender	4.327	1	4.327	0.635	0.433	0.12
	Error (Landing)	163.634	24	6.818			
$\gamma_{TD(TL/L)}$	Gender*	1053.000	1	1053.000	11.098	0.003	0.89
	Error	2277.088	24	94.879			
	Landing	0.019	1	0.019	0.007	0.933	0.05
	Landing × Gender	3.250	1	3.250	1.219	0.280	0.19
	Error (Landing)	63.971	24	2.665			
$\gamma_{TD(TH/TL)}$	Gender	177.970	1	177.970	1.769	0.196	0.25
	Error	2415.068	24	100.628			
	Landing	6.797	1	6.797	1.593	0.219	0.23
	Landing × Gender*	22.231	1	22.231	5.209	0.032	0.59
	Error (Landing)	102.432	24	4.268			
$\gamma_{TD(LC/TH)}$	Gender	4.099	1	4.099	0.013	0.910	0.05
	Error	7534.978	24	313.957			
	Landing	20.188	1	20.188	0.587	0.451	0.11
	Landing × Gender	1.357	1	1.357	0.039	0.844	0.05
	Error (Landing)	825.895	24	34.412			

* Significant main effect or interaction ($p<0.05$)

Table C-4. Univariate tests for the extension ROM variables in the study of landing technique effects.

Measure	Source	Type III SS	df	Mean Square	F	Sig. (p)	Observed Power (a)
Extension ROM variables							
$\gamma_{P(L/S)}$	Gender	1.699	1	1.699	2.030	0.167	0.28
	Error	20.093	24	0.837			
	Landing	2.862	1	2.862	3.854	0.061	0.47
	Landing × Gender	1.422	1	1.422	1.915	0.179	0.26
	Error (Landing)	17.825	24	0.743			
$\gamma_{P(TL/L)}$	Gender	24.647	1	24.647	3.605	0.070	0.45
	Error	164.086	24	6.837			
	Landing	1.357	1	1.357	0.761	0.392	0.13
	Landing × Gender	0.739	1	0.739	0.414	0.526	0.10
	Error (Landing)	42.814	24	1.784			
$\gamma_{P(TH/TL)}$	Gender*	199.685	1	199.685	7.323	0.012	0.74
	Error	654.477	24	27.270			
	Landing*	147.236	1	147.236	30.860	< 0.001	1.0
	Landing × Gender	19.082	1	19.082	3.999	0.057	0.48
	Error (Landing)	114.508	24	4.771			
$\gamma_{P(LC/TH)}$	Gender	36.056	1	36.056	0.296	0.591	0.08
	Error	2924.496	24	121.854			
	Landing*	465.005	1	465.005	9.101	0.006	0.83
	Landing × Gender	18.125	1	18.125	0.355	0.557	0.09
	Error (Landing)	1226.265	24	51.094			

* Significant main effect or interaction ($p<0.05$)

Table C-5. Univariate tests for the L/S kinetic variables in the study of landing technique effects.

Measure	Source	Type III SS	df	Mean Square	F	Sig. (p)	Observed Power (a)
L/S Kinetic variables							
AxF(L/S)	Landing*	98.533	1	98.533	28.603	< 0.001	0.999
	Landing × Gender	4.431	1	4.431	1.286	0.268	0.193
	Error (Landing)	82.677	24	3.445			
$ShF(L/S)_{ant}$	Landing*	10.593	1	10.593	25.316	< 0.001	0.998
	Landing × Gender	2.495	1	2.495	5.962	0.022	0.649
	Error (Landing)	10.042	24	0.418			
$ShF(L/S)_{post}$	Landing*	375.363	1	375.363	74.279	< 0.001	1.000
	Landing × Gender	2.646	1	2.646	0.524	0.476	0.107
	Error (Landing)	121.282	24	5.053			
FlxM(L/S)	Landing*	2.696	1	2.696	23.927	< 0.001	0.997
	Landing × Gender	0.000	1	0.000	0.002	0.961	0.050
	Error (Landing)	2.704	24	0.113			
ExtM(L/S)	Landing*	29.280	1	29.280	59.699	< 0.001	1.000
	Landing × Gender	0.333	1	0.333	0.679	0.418	0.124
	Error (Landing)	11.771	24	0.490			

* Significant main effect or interaction ($p<0.05$)

Table C-6. Univariate tests for the C/T kinetic variables in the study of landing technique effects.

Measure	Source	Type III SS	df	Mean Square	F	Sig. (p)	Observed Power (a)
C/T Kinetic variables							
AxF(C/T)	Gender	1.265	1	1.265	0.105	0.749	0.061
	Error	289.164	24	12.048			
	Landing*	143.989	1	143.989	14.269	0.001	0.952
	Landing × Gender	11.197	1	11.197	1.110	0.303	0.173
	Error (Landing)	242.184	24	10.091			
$ShF(C/T)_{ant}$	Gender	9.883	1	9.883	1.761	0.197	0.247
	Error	134.708	24	5.613			
	Landing	5.617	1	5.617	2.641	0.117	0.345
	Landing × Gender*	9.762	1	9.762	4.591	0.042	0.538
	Error (Landing)	51.034	24	2.126			
$ShF(C/T)_{post}$	Gender*	8.360	1	8.360	5.609	0.026	0.623
	Error	35.774	24	1.491			
	Landing*	21.543	1	21.543	62.879	<0.001	1.000
	Landing × Gender*	5.066	1	5.066	14.785	0.001	0.958
	Error (Landing)	8.223	24	0.343			
FlxM(C/T)	Gender	0.099	1	0.099	0.097	0.758	0.06
	Error	24.521	24	1.022			
	Landing*	8.074	1	8.074	16.232	<0.001	0.971
	Landing × Gender	0.467	1	0.467	0.940	0.342	0.154
	Error (Landing)	11.938	24	0.497			
ExtM(C/T)	Gender	1.688	1	1.688	1.113	0.302	0.173
	Error	36.404	24	1.517			
	Landing*	33.681	1	33.681	47.772	<0.001	1.0
	Landing × Gender	1.596	1	1.596	2.264	0.145	0.303
	Error (Landing)	16.921	24	0.705			

* Significant main effect or interaction ($p<0.05$)

Effects of Knee Joint Muscles Fatigue

Table C-7. MANOVA table for the fatigue indices in the study of knee joint muscles fatigue.

Effect	Roy's Largest Root	F	Hypothesis df	Error df	Sig.(p)	Observed Power (a)
Gender	0.057	0.657	2	23	0.528	0.147
Fatigue*	0.875	10.067	2	23	0.001	0.971
Fatigue × Gender	0.047	0.542	2	23	0.589	0.129

* Significant main effect or interaction (p<0.05)

Table C-8. Univariate tests of within-subjects effects for the fatigue indices in the study of knee joint muscles fatigue.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power (a)
Knee Extensors	Fatigue*	362.525	1	362.525	5.120	0.033	0.584
	Fatigue × Gender	6.582	1	6.582	0.093	0.763	0.060
	Error (Fatigue)	1699.338	24	70.806			
Knee Flexors	Fatigue*	736.509	1	736.509	20.596	< 0.001	0.992
	Fatigue × Gender	40.163	1	40.163	1.123	0.300	0.174
	Error (Fatigue)	858.252	24	35.761			

* Significant main effect or interaction (p<0.05)

Table C-9. MANOVA table for landing variables, touchdown angle variables, extension ROM variables, and L/S and C/T kinetic variables in the study of knee joint muscles fatigue.

Effect	Roy's Largest Root	F	Hypothesis df	Error df	Sig. (p)	Observed Power (a)
Landing variables						
Gender*	1.642	4.104	6	15	0.012	0.88
Fatigue	0.335	0.838	6	15	0.559	0.24
Fatigue × Gender	0.870	2.174	6	15	0.104	0.58
Touchdown angle variables						
Gender*	0.773	3.287	4	17	0.036	0.72
Fatigue	0.216	0.920	4	17	0.475	0.23
Fatigue × Gender	0.110	0.467	4	17	0.759	0.13
Extension ROM variables						
Gender*	1.259	5.351	4	17	0.006	0.92
Fatigue	0.462	1.965	4	17	0.146	0.47
Fatigue × Gender	0.556	2.363	4	17	0.094	0.55
L/S Kinetic variables						
Gender	0.644	2.062	5	16	0.124	0.529
Fatigue	0.748	2.395	5	16	0.084	0.602
Fatigue × Gender*	1.010	3.234	5	16	0.033	0.752
C/T Kinetic variables						
Gender	0.336	1.074	5	16	0.411	0.285
Fatigue	0.476	1.522	5	16	0.238	0.399
Fatigue × Gender*	0.936	2.995	5	16	0.043	0.714

* Significant main effect or interaction ($p<0.05$)

Table C-10. Univariate tests of between-subjects effects for the landing variables in the study of knee joint muscles fatigue.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power(a)
PVGRF	Gender	18.567	1	18.567	1.077	0.312	0.17
	Error	344.679	20	17.234			
$t_{(LP)}$	Gender	0.051	1	0.051	1.111	0.304	0.17
	Error	0.912	20	0.046			
$\theta_{TD(KFA)}$	Gender*	580.417	1	580.417	8.755	0.008*	0.80
	Error	1325.875	20	66.294			
$\theta_{P(KFA)}$	Gender	582.142	1	582.142	1.279	0.271	0.19
	Error	9100.907	20	455.045			
$\theta_{TD(HFA)}$	Gender	743.700	1	743.700	3.087	0.094	0.39
	Error	4817.829	20	240.891			
$\theta_{P(HFA)}$	Gender	20.202	1	20.202	0.062	0.807	0.06
	Error	6561.283	20	328.064			

* Significant main effect or interaction ($p<0.05$)

Table C-11. Univariate tests of between-subjects effects for the touchdown angle variables in the study of knee joint muscles fatigue.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power (a)
$\gamma_{TD(L/S)}$	Gender	83.277	1	83.277	0.535	0.473	0.11
	Error	3111.027	20	155.551			
$\gamma_{P(TL/L)}$	Gender*	1201.868	1	1201.868	9.729	0.005	0.84
	Error	2470.578	20	123.529			
$\gamma_{P(TH/TL)}$	Gender	405.650	1	405.650	3.730	0.068	0.45
	Error	2175.019	20	108.751			
$\gamma_{P(LC/TH)}$	Gender	19.032	1	19.032	0.054	0.819	0.06
	Error	7090.599	20	354.530			

* Significant main effect or interaction ($p<0.05$)

Table C-12. Univariate tests of between-subjects effects for the extension ROM variables in the study of knee joint muscles fatigue.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power (a)
$\gamma_{P(L/S)}$	Gender	2.401	1	2.401	3.567	0.074	0.44
	Error	13.463	20	0.673			
$\gamma_{P(TL/L)}$	Gender	18.888	1	18.888	3.295	0.085	0.41
	Error	114.659	20	5.733			
$\gamma_{P(TH/TL)}$	Gender*	328.901	1	328.901	7.437	0.013	0.74
	Error	884.457	20	44.223			
$\gamma_{P(LC/TH)}$	Gender	17.388	1	17.388	0.084	0.776	0.06
	Error	4162.524	20	208.126			

* Significant main effect or interaction ($p<0.05$)

Table C-13. Univariate tests of within-subjects effects for the kinetic variables of L/S junction in the study of knee joint muscles fatigue.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power (a)
AxF(L/S)	Fatigue	0.013	1	0.013	0.016	0.900	0.052
	Fatigue × Gender*	5.186	1	5.186	6.529	0.019	0.681
	Error (Fatigue)	15.887	20	0.794			
$ShF(L/S)_{ant}$	Fatigue	0.003	1	0.003	0.024	0.877	0.053
	Fatigue × Gender	0.209	1	0.209	1.884	0.185	0.258
	Error (Fatigue)	2.213	20	0.111			
$ShF(L/S)_{post}$	Fatigue	11.637	1	11.637	12.694	0.002	0.923
	Fatigue × Gender*	12.752	1	12.752	13.910	0.001	0.944
	Error (Fatigue)	18.335	20	0.917			
FlxM(L/S)	Fatigue	0.025	1	0.025	0.323	0.576	0.084
	Fatigue × Gender	0.000	1	0.000	0.003	0.961	0.050
	Error (Fatigue)	1.571	20	0.079			
ExtM(L/S)	Fatigue	1.047	1	1.047	8.029	0.01	0.769
	Fatigue × Gender*	1.036	1	1.036	7.943	0.011	0.765
	Error (Fatigue)	2.609	20	0.130			

* Significant main effect or interaction ($p<0.05$)

Table C-14. Univariate tests of between-subjects and within-subjects effects for the kinetic variables of C/T junction.

Measure	Source	Type III Sum of Squares	df	Mean Square	F	Sig. (p)	Observed Power (a)
AxF(C/T)	Fatigue	3.734	1	3.734	3.048	0.096	0.383
	Fatigue × Gender*	15.864	1	15.864	12.947	0.002	0.928
	Error (Fatigue)	24.507	20	1.225			
$ShF(C/T)_{ant}$	Fatigue	0.018	1	0.018	0.014	0.907	0.051
	Fatigue × Gender	0.174	1	0.174	0.138	0.715	0.064
	Error (Fatigue)	25.305	20	1.265			
$ShF(C/T)_{post}$	Fatigue	1.275	1	1.275	5.030	0.036	0.569
	Fatigue × Gender	0.039	1	0.039	0.154	0.699	0.066
	Error (Fatigue)	5.068	20	0.253			
FlxM(C/T)	Fatigue	0.104	1	0.104	0.533	0.474	0.107
	Fatigue × Gender	0.015	1	0.015	0.075	0.787	0.058
	Error (Fatigue)	3.898	20	0.195			
ExtM(C/T)	Fatigue	1.283	1	1.283	6.133	0.022	0.654
	Fatigue × Gender	0.874	1	0.874	4.175	0.054	0.494
	Error (Fatigue)	4.185	20	0.209			

* Significant main effect or interaction ($p<0.05$)

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BIOGRAPHICAL SKETCH

Soo-An Park was born on January 19, 1965 to his parents Seung-Jae Park and Jung-Nim Kang in Seoul, Korea. He is married to Hyunhee Kwon, and has one son, David Joonsuh. He also has one younger sister, Eun-Hee and one younger brother, Soo-Min. Soo-An lived in Seoul, Korea throughout his childhood until he became an orthopaedic surgeon. He enrolled in the Medical College in the Catholic University of Korea, and worked as an intern and a resident of orthopaedic surgery at Kangnam St. Mary's Hospital after graduation.

During his residency, Soo-An was interested in the study of spine and sports medicine. His interest in the field of spinal surgery led him to pursue a Spine Research Fellowship under the guidance of Hansen A. Yuan, M.D. in SUNY-Upstate Medical University. A quest for interesting spine research and an interest in human performance in the sport led him to the Biomechanics Laboratory at the University of Florida and to doctoral work. New academic information in biomechanics and his clinical experience in orthopaedic spine surgery provided the foundation for his dissertation.

Soo-An will continue his research work as he has accepted a 1-year postdoctoral fellowship in the Department of Orthopaedic Surgery at SUNY-Upstate Medical University. He will resume his clinical work in 2008 when he returns to the Department of Orthopaedic Surgery at Asan Medical Center of Ulsan Medical University in Seoul, Korea, where he will specialize in spinal surgery.