APPLICATION OF JERK ANALYSIS TO A REPETITIVE LIFTING TASK IN PATIENTS WITH CHRONIC LOWER BACK PAIN

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Patients with chronic lower back pain (CLBP) typically demonstrate different biomechanics than healthy controls during a lifting task. Motion differences in a repetitive lifting task have been described previously using differences in the timing of body angles changes during the lift. These timing changes rely on small differences of motion and are difficult to measure and to interpret. The purpose of this study is to evaluate shoulder jerk (rate of change of acceleration) in a repetitive lifting task as a parameter to detect differences of motion between controls and CLBP patients and to measure the impact of a rehabilitation program on jerk. The jerk calculation proved to be a noisy measure since jerk is the third derivative of position, and a simulation study was performed to evaluate smoothing methods to provide the best estimates of the third derivative. Woltring's generalized cross-validation spline produced the best estimates and was fit to subject data. Derivatives were calculated using differentiation of the spline coefficients, and root-means-square (rms) amplitude of jerk was used for comparison. Lifts were divided into phases of early, middle or late based on the number of repetitions completed by the subject. Average values of rms jerk during a lift were computed at each of the task phases. Significant group differences were found for rms jerk. CLBP patients were found to perform lifts with lower jerk values than controls and as the task progressed, rms jerk increased for both groups. A group-by-phase interaction was significant. After completion of a rehabilitation

program, CLBP patients performed lifts with greater rms jerk. In general, patients performed lifts with lower jerk values than controls, suggesting that pain impacts lifting style.

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PREFACE

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1.0 INTRODUCTION

Lower back pain is the leading cause of disability in people under the age of 45 years and is expected to afflict between 60% and 90% of individuals during their lifetimes [1]. A survey conducted from August 2001 to July 2002 using data from the American Productivity Audit estimated that back pain results in a $19.8 billion per year loss due to pain-related lost productivity time [2]. Approximately 76.6% of this cost is due to health related reduced performance that occurs while employees are at work [2]. The number of workdays of reduced performance and subsequent lost productivity due to pain varies by condition, with back pain accounting for 69.7% [2].

The precise etiology of back pain is unknown. Several hypotheses including muscle incoordination during fast motions, muscle fatigue with repeated exertions, and cumulative disc degeneration due to high, repeated disc compression forces have been proposed [3]. Compressive forces are imposed on the lower back through motions such as lifting and bending. During lifting, trunk muscle contractions, load position, back curvature, body posture and intra abdominal pressure are among the many factors affecting the compressive forces on joints [4].

Due to the increasing cost of health care and lost productivity associated with lower back pain, several researchers have focused on describing differences in flexibility and in the motion between the hip and knees during lifting. In a flexion task conducted by Lavivierie et al., chronic lower back pain patients (CLBP) demonstrated less lumbar flexion than pain-free controls [5]. Chronic lower back pain is a condition in which patients experience pain in the lower back for more than six months. Sparto et al found that fatigue resulted in a reduction in joint torques and lifting force during a repetitive lifting task [6]. Fatigued subjects showed a decrease in hip and knee motion and an increase in spinal flexion. As the task progressed, knee extension preceded hip extension. Scholz found relatively continuous changes in a measure of coordination during squat lifts as a result of increased load [7]. Boston et al. determined a coordination index to characterize differences of motion between the hip and knee angles between CLBP patients and pain-free controls [8]. Pain-free controls were found to move the hip and knee asynchronously when they initiate the lift, but the hip and knee angles tended to reach full extension simultaneously at the end of the lift. The investigators interpreted this lifting style as a coordinated ending. CLBP patients used a lifting pattern in which the hip and knee ended motion at different times. This lifting style was described as an uncoordinated ending.

These studies have suggested that lifting patterns describing inter-segmental motion may be an important characteristic to differentiate CLBP patients from pain-free controls. However, inter-segmental motion relies on small differences between the parameters that describe the hip and knee angles as a function of time and can be difficult to interpret. For instance, the coordination index described by Boston et al described motion as a function of time using the parameters of midpoint and risetime. The midpoint is the time at which half the angle motion has occurred and the risetime is the time required for the angle to decrease from 88% to12% of the total change in angle [8]. Relative motion between the hip and knees was described by the difference between the hip and knee midpoint and the hip and knee risetime. The risetime and midpoint differences are small parameters that are highly susceptible to noise. The noise increases the variance within the parameters, making differences between groups or changes over time difficult to detect. A correlation between the parameters was found for controls but not for patients. This suggests that patients and controls move differently during a lifting task. However, the variability within these parameters was too great to detect differences in individual subjects.

A search for a more robust measure than the coordination index to differentiate lifting patterns between CLBP patients and controls has lead to jerk, which is defined as the rate of change of acceleration or the third derivative of position. Flash and Hogan proposed jerk as the objective function of a motion control model [9]. The minimum jerk model proposes that the central nervous system plans a trajectory of a movement in a manner that would minimize jerk (changes in acceleration) and maximize smoothness of motion [10]. Flash and Hogan validated the model on movements involving planar two joint arm motion. With practice and learning of a movement, subjects produced motion with less jerk resulting in smoother trajectories of motion [9].

Applying the minimum jerk model to a lifting task would suggest that motion becomes smoother as the task progresses. However, some studies have found that multi-segmental tasks do not obey the minimum jerk model because jerk is increased with practice and learning [11]. A jerk lift requires large changes of acceleration in motion, which could translate to large changes of forces on the body especially the lower back. These forces may cause pain in CLBP patients. Patients are more likely to utilize a lifting style in which their pain is minimized resulting in a smoother, lower jerk lifting style. Thus, I hypothesize that patients will employ a motion control strategy of performing smoother, lower jerk lifting style. Control subjects are more flexible and not limited by pain, making them capable of sustaining large changes of force. Controls are hypothesized to perform lifts that have a large magnitude of jerk.

In this project, root-mean-squared (rms) amplitude of jerk was calculated for data collected from a repetitive lifting task involving CLBP patients and pain-free controls. Jerk was used as a parameter to detect lifting pattern differences between groups and the impact of treatment on lifting patterns. The subject data sample used was the same subject data utilized previously by Boston et al [8][12] to describe inter-segmental motion. An advantage of using the same data is that lifting pattern differences between the groups have already been established. The data set will provide a test to determine whether rms jerk can detect lifting pattern differences that exist between controls and CLBP patients. Parameters previously used, such as starting posture and lift duration [13], will be compared to rms jerk to provide further description of lifting patterns and motion control strategies.

Calculation of jerk proved to be non-trivial since jerk is a higher order derivative. Derivatives are highly sensitive to certain types of errors in the measurement and therefore smoothing and/or filtering techniques must be utilized [14]. Review of the literature suggests that noise varies depending on the biomechanical task in which each separate task can require a different optimal smoothing criterion. Giakas and Baltzopoulos tested several smoothing methods on walking data at various noise levels and found that no one optimal solution or automatic method to filtering biomechanical data existed [15]. Each of the methods had advantages and disadvantages that depended on the derivative calculated and the noise level. Careful evaluation of each smoothing method on synthetic data must first occur in order to achieve the best estimate of the derivatives.

The primary hypothesis of this project is that jerk, with the application of a smoothing technique, will be able to distinguish lifting pattern differences between CLBP patients and pain-free control subjects during a repetitive lifting task and do so more robustly then previous measure. These differences are hypothesized to result in CLBP patients' use of a lifting style that produces a lower magnitude of jerk when compared to pain-free controls. Secondly, CLBP patients will have higher jerk magnitude after completion of rehabilitation program.

The organization of this thesis is as follows. Chapter 2 details previous research studies that examined differences in muscle activation, lifting techniques and flexibility between CLBP patients and pain-free controls. Also discussed in this chapter are studies utilizing jerk as a parameter to describe motion and studies testing smoothing methods on biomechanical data. Chapter 3 describes a simulation study to compare several smoothing methods. Chapter 4 describes the experimental methods, experimental protocol, instrumentation, parameter calculation, and statistics methods. Differences of jerk due to group, treatment and changes over time are presented in Chapter 5. Chapter 6 discusses the use jerk as a parameter and interprets the results of jerk obtained in Chapter 5.

2.0 BACKGROUND

This chapter reviews previous work on lifting, the use of jerk as a biomechanical measurement, and methods for smoothing data.

2.1 LIFTING STUDIES

The purpose of this section is to present studies describing motion changes during a repetitive lifting task with emphasis on lifting pattern differences between CLBP patients and pain-free controls. Repetitive lifting tasks have been reported to initiate fatigue, which causes distinct changes in body segmental motion, and flexibility in pain-free control subjects. CLBP patients have demonstrated differences during lifting in body motion when compared to pain-free controls. These motion differences include flexibility, spinal loading, lifting speed and segmental motion.

Dolan tested whether repetitive bending and lifting tasks lead to fatigue in the back muscles in subjects with no history of back problems [16]. Fatigue of the back muscles were measured by EMG analysis of the erector spinae muscles and defined as a percent reduction of the baseline static back strength measurement specific to the subject. The repetitive lifting task required the subjects to lift and lower a 10 kg disc 100 times from the floor to waist height at a constant pace determined by the subject. Lumbar flexion was measured during the task and compared to a range of lumbar flexion obtained for a particular subject before the start of the task. Lumbar curvature was related to the peak bending moment of the spine. As the task progressed, spinal bending increased as the estimated peak extensor moment decreased and lifting speed increased. This reduction in the peak extensor moment indicated a change from a squat lift to a torso lift. A torso lift is a lift that begins with the back bent and the knees at approximately full extension, producing less compression but greater shear forces on the lumbar spine. A squat lift is a lift that begins with the knees fully bent and the back straight, producing less shear force but greater compressive force on the lumbar spine. A freestyle lift is classified as a lifting style that is a combination of a squat and torso style lift [12]. Bending moment on the osteoligamentous spine was increased as indicated by the increase of flexion of the lumbar spine. Subjects demonstrated greater flexion that resulted in larger forces acting on the ligaments of the spine at the end of the task and showed measurable fatigue of the erector spinae muscles [16].

Bonato et al also studied fatigue of the back muscles during a repetitive lifting task [17]. Several pain-free male subjects lifted a box from midshank to waist level at a pace of 12 lifting cycles per minute for a total of 5 minutes. Electrodes placed on the superficial back muscles recorded EMG signals to determine if fatigue was induced by this task. Angle changes of body segments were also measured using markers placed on anatomical landmarks to compare changes in the muscle activity to changes in motion patterns. Analysis of the EMG signals and biomechanical parameters were performed on 6 lifting cycles at the beginning and end of the task. Six of the 9 subjects enrolled in the study showed a relationship between changes in EMG characteristics and biomechanical parameters. When instantaneous median frequency of the

EMG signal decreased, motion in the upper and lower limb changed. This finding suggested that the body adapts motion due to muscle fatigue at the lumbar region.

The fatigability of the lumbar paraspinal and the gluteus maximus muscles in CLBP and healthy controls was explored by Kankaanpää et al [18]. Each subject in the study performed a seated dynamic back extension task under a resistant load of 15-20 kg, during which EMG signals of lumbar paraspinal and gluteal muscles were recorded. The subjects were required to perform the task until exhausted. The gluteus maximus muscle was found to fatigue faster in CLBP subjects than in controls. However paraspinal muscle fatigue was similar in both groups. The gluteus maximus is a strong hip extensor and tightly coupled to the lumbar paraspinal muscles, allowing for transfer of load from the spine to the lower extremities. Kankaanpää suggested that general deconditioning associated with CLBP patients may explain the greater fatigue of the hip extensor muscles.

Laviviere et al quantified a distinction in flexibility between pain-free controls and CLBP patients for a weighted bending motion [5]. Each subject was required to perform an unweighted and weighted lateral bending and forward flexion motion. The total weighted and unweighted motion was performed continuously and designated as one cycle. The subject performed the cycle 3 times, with a 2-minute rest period between cycles. Lateral bending showed no significant difference between the groups. The flexion task did show significant group difference regarding the contribution of the thoracic and lumbar regions. Patients had a higher thoracic and lower lumbar contribution than controls. These results suggest that CLBP patients might have attempted to protect their lumbar passive tissues by using less lumbar flexion. The increase of thoracic contribution may have been compensation for the limited lumbar function. Laviviere et al. suggested CLBP patients compensate for the limited function by using larger thoracic flexion to reach the floor with their hands during the unloaded task [5].

A similar study performed by McIntyre examined flexion and extension motion differences between low back pain and control subjects [19]. The repetitive dynamic flexion and extension task required subjects to bend forward against a resistant load while constrained in a position where the knees were kept straight. Each subject worked against a resistant load set to 50% of his or her maximum voluntary flexion torque. Pace of movement and range of motion were controlled by the subject, and each subject performed the task for 120 seconds or until exhausted. Low-back motions between the groups differed. Lower back pain subjects flexed and extended at a lower velocity than control subjects and performed the task using a smaller range of motion. Lower back pain subjects restricted motion to a limited range about the neutral vertical position, possibly to avoid an increase in torque on the back.

Marras et al. investigated spinal loading through EMG analysis of CLBP subjects compared to controls during static and dynamic lifting tasks [20]. The subjects were required to complete two tasks: a static exertion and a dynamic lifting task. The static exertion involved sagittal extension against three different resistant loads. During the extension, posture was controlled by fixing the subject's pelvis to a structure that measured trunk moment. The dynamic lifting task required subjects to lift four different weights, each starting from six different origins. Trunk muscle activity, trunk kinematics and trunk kinetics were obtained from EMG activity of 10 trunk muscle sites and used in a three-dimensional model to determine spinal loading. Lower back pain subjects experienced 26% more compression and 75% more shear than control subjects for the controlled posture static exertion task. In the dynamic task, low back pain subjects reduced trunk moment through less flexion and motion. This was not an effective lifting strategy, since the low back pain subjects were found to have higher spinal compression, especially for lifts below waist level due to increase muscle coactivity.

Bush-Joseph et al and Buseck et al. found that an increase in lifting speed produced an increase in the L5/S1 moment [21] [22]. In the Bush-Joseph et al. study, healthy subjects lifted a 150N box using three different lifting postures (back, leg and free-style lift) at three speeds (slow, normal and fast). Subjects performing a torso lift at slow and normal lifting speeds had 15-20% lower L5/S1 moment than when performing a squat lift and 20-25% lower L5/S1 moment than when performing a free-style lift. In the Buseck et al. study, healthy subjects were required to lift a box varying in weight from 50 to 250 N at normal and fast speeds using a leg lift and/or freestyle lift. The peak L5/S1 moment was found to increase linearly with increasing load and higher moments occurred at faster lifts for all weights.

Scholtz characterized coordination through observation of differences in motion between the knee and the lumbar spine [7]. Five male subjects were required to lift loads equivalent to 0, 15, 30, 45, 60 and 75% of their voluntary maximum lifting capacity for four trials per load. Subjects were instructed to perform a squat style lift at a comfortable lifting speed. Results showed that the magnitude of the load dictated the coordination of the knee and lumbar spine. The lumbar spine was found to extend later with respect to the cycle of knee motion as the load became heavier.

Sparto et al. also observed segmental motion disparities in a repetitive lifting task involving pain-free male subjects [6]. Fatigue was induced in this task by requiring each subject to lift at his maximal rate until the subject was unable to continue. The subjects lifted a box from midshank to waist level as quickly as possible. Motion in two dimensions at the ankle, knee, hip, elbow, shoulder and sagittal flexion and extension were measured. The subjects showed an overall decrease in hip and knee motion and an increase in spine flexion as they fatigued, indicating a change from a squat lift to a torso lift [6]. Knee extension preceded hip extension during the task, and subjects showed a reduction in lifting force and joint torque as the task progressed.

Similar differences in motion between adult CLBP patients and pain-free controls have been found in a repetitive lifting study by Boston et al. [8]. This study described a coordination index derived from hyperbolic tangent curves that were fit to the hip and knee angle data as functions of time. Each curve was described by four parameters: starting angle, ending angle, risetime and midpoint time. The risetime was defined as the time required for the angle to decrease from 88% to12% of the total change in angle. The midpoint time defined the time after the beginning of the lift at which the hip or knee angle had completed half of its range of motion. The difference between the hip and knee risetime and the difference between the hip and knee midpoint described relative movement of the two joints. The difference is zero when a subject moves the hip and knees at the same speed, resulting in synchronous motion. In this case, both angles reach the midpoint at the same time. A negative midpoint difference occurs when the hips straighten before the knees. A positive risetime difference indicates that the knee angle is changing faster than the hip angle producing a longer hip risetime than knee risetime.

Pain-free controls were found to move the hip and knee asynchronously when they initiated the lift, but the hip and knee angles reach full extension simultaneously at the end of the lift. Controls either moved the knees earlier and the hip moved faster or the hip moved earlier and the knees moved faster in order to end together. For this lifting style, the midpoint and risetime differences are opposite in sign and the hip and knee motion produce a coordinated ending. CLBP patients used a "guarded" lifting pattern in which the hip and knee moved synchronously when patients initiated the lift but the hip and knees angles finished motion at different times. In this lifting style, the midpoint and risetime differences are the same sign and the hip and knee are not coordinated at the end of the lift. The guarded lifting style can result from contracting both agonist and antagonist muscles of a joint and has been suggested as a mechanism to avoid or minimize pain during movement [23].

Boston et al. extended the coordination index study to assess the impact of rehabilitation on CLBP subjects. The coordination index showed that CLBP patient's pre- and post-treatment index changed significantly, showing more coordinated endings between the hip and knee post-treatment. The post-treatment coordination indices were not significantly different from those observed in controls. The test-retest reliability of the coordination index was found to be high at 0.76 [12].

Similar motion pattern differences were found in several other parameters in this study. These parameters are starting posture, lift duration and a work index. The starting posture measures whether the subject performed a torso lift or a squat lift. CLBP patients pre-treatment were found to use more of a squat lift than controls, and post-treatment patients showed a greater squat lifting style. Lift duration is the time required for the subject to perform a lift. Control subjects performed lifts faster than CLBP patients pre- and post-treatment. Post-treatment testing resulted in decreased lift duration of CLBP patients when compared with pre-treatment testing. Work index was defined as the number of lifts performed multiplied by the weight lifted. The work index was greater for control subjects than CLBP patients. Treatment showed a 71% increase in the work index from pre-treatment values. However the post-treatment values never approached the work indices of control subjects [12].

A second paper by Rudy et al. investigated differences over time of the pre-treatment parameters [13]. Lifts were grouped as early, middle, or late phase, based on the individual subject's number of repetitions, to minimize the effects of wide variations among subjects in the number of lifts performed. Repeated-measures ANOVA determined differences between experimental group and changes over time. Starting posture changed over time for CLBP patients demonstrating a greater knee angle as the task progressed from the early to the middle phase. Control subjects consistently produced greater hip flexion than patients throughout the task. The hip-knee midpoint time difference showed dissimilar changes over task time for controls compared to patients. Control subjects increased hip and knee midpoint as the task progressed but patients increased midpoint from early to middle phase and then stabilized from the middle to the late phase. Lift duration demonstrated the same changes over task time for both groups, in which speed increased as the task progressed from early to middle phase and then speed stabilized. Controls performed faster lifts than patients for all phases. The coordination index could not be used in the early, middle, late analysis, since the coordination index produced a single parameter for the entire task.

Review of the lifting studies showed that fatigue caused subjects to increase flexion, reduce lifting force and caused subjects to change from a squat lifting style to a torso lifting style. CLBP patients demonstrated greater fatigueability of the hip extensor muscles and performed lifts with less flexion than controls. The limited flexion caused greater compression and shear forces onto the lower back during lifting. Patients performed lifts with uncoordinated inter-segmental motion, which resulted in the hip and knee joints ending motion at different times. Control subject's hip and knee joints moved asynchronously at the start of the lift, but the joints coordinated motion to end the lift together. Patients performed lifts slower than controls and used guarded lifting style by coactivating muscles.

2.2 JERK

Jerk (the time derivative of acceleration) was postulated by Flash and Hogan to describe the smoothness of planar arm movements [9]. A mathematical model to describe voluntary arm movements was formulated and tested with subject data. The model involved dynamic optimization in which a criterion function was used to describe the objective of movement [9]. The criterion function is that the central nervous system plans to maximize smoothness of motion by minimizing jerk. The model was solved to obtain a predicted displacement trajectory to minimize jerk as a 5^{th} order polynomial in both x and y paths. The velocity profile, predicted by the model, for hand paths is a bell-shaped curve. The study determined the arm trajectory that produced bell-shaped velocity profiles for each of the tasks and compared these trajectories to subject data to validate the minimum jerk model.

Subjects were required to move a two-link mechanical manipulandum to targeted positions under four different conditions. The first condition required each of the subjects to move the manipulandum to an illuminated target position. The subject was not aware of the target location until the target was illuminated. The second condition required the subject to move at a designated movement speed. Targets were illuminated along a path at the desired pace and the subject was instructed to move accordingly. The third condition generated paths for the subjects to follow from start to target position. The final task involved maneuvering around an obstacle to the target position. In all of the tasks, the subject's shoulder was constrained. The results showed that all the observed motion trajectories were consistent with predicted minimum jerk trajectories.

Schneider and Zernicke utilized jerk to quantify whether learning would produce smoother rapid arm movements [24]. Subjects were required to lift a plate from a low target to high target and then back to the low target. A barrier had to be circumnavigated between the targeted positions. The subjects completed 100 practice trials as quickly as possible with a small rest period between each trial. After a rest period, the subjects repeated the experiment at their slowest, mid-range and fastest motion exhibited during the practice trials. Motion was recorded by high-speed cine film. Markers were placed on the subject's shoulder, elbow, wrist and third metacarpophalangeal joint and derivatives were calculated from a fit of Woltring quintic spline [14] to the position data. Results of the practice trials showed that movement times decreased and mean-squared jerk (jerk-cost) increased as the task progressed. Schneider and Zernicke suggested that the increase in jerk-cost is due to the increase in speed required by the task since subjects were instructed to perform as quickly as possible. Learning was quantified by comparing identical movement times of practice trials with after-practice trials. Jerk-cost was found to decrease for identical duration movements in after-practice trials, indicating an increased smoothness of learned movements.

Hreljac observed similar results for a lower limb obstacle avoidance task [25]. The obstacle avoidance task required the subjects to step over two parallel obstacles as quickly as possible. The right foot of the subject was tracked to describe jerk of the lower limb during the task. The data were filtered using a 4^{th} order Butterworth filter. Derivatives were calculated using finite differences and filtered at optimal frequencies described by the Wells and Winter residue method [26][27]. As the subjects practiced the task, jerk-cost values decreased along with movement time. The correlation between movement time and decreased jerk-cost values indicate that moving more rapidly was associated with moving more smoothly.

Jerk has been used successfully to distinguish motion difference between groups with varied ability and experience. Hreljac compared jerk-cost during a running and fast walking task between a group of competitive runners and non-running athletes [28]. Subjects walked or ran on a treadmill for 15 minutes, and motion was videotaped for the final 10 minutes. The video tracked markers located on the lateral aspect of the heel of the left shoe and mid-sole for three consecutive strides.

This study revealed that differences in the derivatives of motion between the groups became greater as the derivative order became higher. The end-point jerk-cost results showed that competitive runners had lower jerk values for the walking task than the non-runner athletes. In the running condition, only the stride and swing phase of the motion showed significant differences with competitive runners producing smoother (lower jerk) motion. Hreljac suggested that if all subjects are assumed to minimize jerk, competitive runners may exhibit higher level of coordination in walking and running movements, since runners were more successful than non-runners in producing smoother (lower jerk) motion [28].

Young and Marteniuk reported a contradictory finding of jerk during a repetitive kicking task [11]. The task required subjects to perform a kicking motion with a weight attached to the ankle. Subjects kicked from a start position to a targeted position while navigating over a barrier and were asked to perform the kick within 400ms. Each subject performed 15 blocks of 16 trials with knowledge of their motion time provided after each trial, followed by one block with no knowledge of motion time. Motion was tracked at the hip, knee, ankle, and the 5th metatarsal head of the kicking leg, which was designated as the end-effector for calculation of the rms value of jerk. Derivatives were calculated using finite differences and the residues method described by Wells and Winter. Learning was considered to occur after the first block of trials. The results showed that when comparing kicking movements performed at similar movement times, rms value of jerk increased with learning. Further results demonstrated that movements performed at different speeds and path distances could produce similar rms jerk values. Young and Martenuik concluded that acquisition of the kicking movement did not lead to the production of smoother movement trajectories, as jerk values late in learning were rarely comparable to the lowest jerk value observed early in learning [11].

Puniello et al. used jerk to characterize the differences in lifting motion between functionally limited elderly subjects and elderly controls [29]. Subjects lifted a box from floor to knee height. A reflective marker was placed on the box to track its displacement and jerk was calculated by differentiating the displacement data of the box. The displacement data were filtered using a mean boxcar-smoothing window of 0.15s. Derivatives were then calculated and the second derivative was smoothed with a mean boxcar window of 0.1s. The smoothing parameters were found using fast Fourier transformation analysis. The velocity profiles of the box showed two separate profiles: unimodal and bimodal. The bimodal velocity profile was typically observed in the subjects with less knee extensor strength. Puniello concluded that the weaker subjects used lower jerk by breaking up the lift into two distinct movements resulting in a bimodal velocity profile. Stronger subjects merged the entire lift task into one motion producing an unimodal velocity profile. Weaker subjects not only had lower magnitudes of jerk but also less trunk momentum than stronger subjects, suggesting that weaker subjects use a more conservative lifting strategy.

2.3 SMOOTHING

As described in the previous section, different smoothing methods were used by different investigators to calculate jerk. The two most common methods are Woltring's generalized cross-validation splines and the residual analysis of Wells and Winter. In this section, both of these smoothing methods are reviewed and additional studies that evaluated these methods and other smoothing methods are described.

Woltring developed a program to estimate derivatives by using spline functions to smooth biomechanical displacement data [14]. The amount of smoothing is dependent on the type of spline used and the smoothing parameter μ. For $\mu=0$, the resultant spline will be an interpolating spline. Interpolating splines are composed of local polynomials of degree greater than or equal to 2m-1, and these polynomials are piecewise continuous at the knots up to and including the (2m-2)nd derivative [14]. The parameter m refers to the order of the spline, where m = 2 refers to a cubic spline, m = 3 refers to a quintic spline, m = 4 refers to a hepatic spline etc. For μ greater than zero, the resultant spline is a natural spline. Natural splines are similar to interpolating splines with the exception that these splines impose a zero boundary condition. The parameter μ must be selected to prevent oversmoothing of the data. Woltring used the cross-validation method of Craven and Wabha to find the optimal value of μ [30]. The method fits a spline function to all data except for the measurement at time t_j (where j = 1,2…n). A fitting error between the spline and the raw data is evaluated at t_j, and μ is defined as the minimum root-mean-squared sum of these fitting errors [30].

Natural splines require that the (2m-2)nd derivative vanishes at the boundary of the data which may lead to end effect errors. Vaughan quantified the second derivative of a free falling golf ball using several different smoothing algorithms, including quintic and cubic splines [31]. The smoothing parameter μ was determined by trial-and-error. The cubic spline fit required the second derivative to be non-zero at the boundary, which was not acceptable for these data. The natural quintic spline provided the best estimate of acceleration since the natural quintic spline required that the third derivative vanish at the boundary.

Woltring used synthetic data to characterize the influence of violating the boundary conditions [14]. The synthetic data were fit to a natural quintic spline and then differentiated. Two conditions were tested: a zero-boundary 3^{rd} derivative and a constant 3^{rd} derivative. The latter condition violated the boundary constraint. The zero-boundary 3^{rd} derivative data resulted in a good estimate of the 3^{rd} derivative and lower order derivatives. The constant 3^{rd} derivative data showed large variability throughout the data, producing an unusable estimate of the 3^{rd} derivative. Woltring concluded that violation of boundary conditions results in artifacts throughout the data and not merely at the ends. For these reasons, Woltring recommended that half the spline order should be greater than the highest derivative sought to avoid artifacts. This recommendation would require that cubic splines be used for 1^{st} derivative, quintic splines for the 2^{nd} derivative, hepatic splines for 3^{rd} derivative etc.

As an alternative method to smooth data before differentiating, Wells and Winter designed a method to find an optimal cut-off frequency for a Butterworth filter using residual analysis [26] [27]. The residual analysis involved filtering data at a certain cut-off frequency and then determining the rms amplitude between the filtered signal and the unfiltered signal. This process was repeated for all possible cut-off frequencies (0 Hz to half the Nyquist frequency), and the rms residuals were plotted versus frequency. The plot has three main regions: (1) region where the kinematic signal predominates, (2) region where signal and noise have similar magnitudes, and (3) the region that is primarily noise. Wells and Winter proposed a method to identify regions 1 and 3 and defined the optimal cut-off frequency as the point where the noise component is five times the signal component.

Hreljac tested several smoothing methods on synthetic data to find the best estimates of the third derivative [28]. These methods were Fourier transforms, quintic splines and single, double, and triple application of a digital filter with residual analysis to determine the optimal cut-off frequency. Extra points were added to the beginning and ending of the signal before applying the smoothing methods to minimize error at the boundaries of the signal. The smoothing method that produced the lowest rms deviation of the third derivative between the known function and the known function plus random noise was used. Results showed that double application of the digital filter with 30 extra frames added to the data provided the best estimate of the 3rd derivative. The filter was applied to the second derivative and the third derivative at different optimal cut-off frequencies.

Vint and Hinrichs evaluated endpoint error in smoothing and differentiation produced by four different methods: Butterworth filter, cubic spline, quintic spline and Fourier series [32]. Endpoint error was described as erratic behavior at the beginning and end of the computed acceleration. Smoothing was applied to angular displacement data with a known acceleration function. The data set was divided into three subsets corresponding to peaks in the known acceleration creating endpoints of high magnitude. Derivatives were computed by differentiation of coefficients for the splines and Fourier series methods and by finite differences for the Butterworth filter method.

Before the application of smoothing and differentiation, the data were padded with 20 points to construct three conditions: data without padding, data plus linear extrapolation padding, and data with padding produced by reflection. Cut-off frequencies of the Butterworth filter and smoothing parameter of the splines were obtained by minimizing the rms of the residuals between the calculated acceleration and the known acceleration. For the Fourier series, the method described by Anderson and Bloomfield [33] was used to provide smoothing. To evaluate endpoint error, the first and last 10 data points of the original data were not smoothed. This avoided confounding effects of the endpoint error, since error within the middle of the data set will affect the endpoints.

After the data were smoothed and differentiated, the rms values between the calculated acceleration and the known acceleration of the unsmoothed data points determined the endpoint error. The quintic spline without padding demonstrated the lowest rms endpoint error. Linear extrapolation of Fourier series, cubic spline, quintic spline and Butterworth filter all demonstrated low error. Linear extrapolation performed better for all methods than padding by reflection. The Butterworth filter performance was highly variable when not padded.

Giakas and Baltzopoulos evaluated the performance of six smoothing techniques on gait [15]. The smoothing techniques were power spectrum assessment, generalized cross-validation spline, least-squares cubic spline, regularization Fourier series, regression analysis and residual analysis. Derivatives for the generalized cross validation spline and the least-squares cubic spline were found from differentiation of the polynomials. Derivatives for all the other techniques were calculated with first-order finite differences.

The original known signals were walking data using 8 three-dimensional displacement markers, which created 24 signals. From these known signals, random noise with 30 noise amplitudes each were added to the known walking signals, creating a total of 1044 noisy signals. The signals were smoothed and differentiated to obtain velocity and acceleration. RMS error between the known derivatives and calculated derivatives determined the performance of the smoothing techniques. The results revealed that performance of each smoothing technique depended on noise level and rms error tended to increase with noise level. Also, for all techniques, derivative calculation affected performance. Optimal smoothed displacement data did not guarantee optimal smoothed velocity and acceleration. No one technique was optimal for all conditions. The best results were obtained from the power spectral estimation, least-squared cubic splines and the generalized cross-validation spline.

Giakas and Baltzopoulos used the same data to determine an optimal digital filtering sequence for noisy biomechanical data [34]. This study used a recursive second-order Butterworth filter and extrapolated at both ends of the data using a reversed mirror method. Finite differences were used to calculate the higher order derivatives. The optimal cut-off frequency was determined by minimizing the rms error between the reference signal and the filtered signal for every derivative. The first procedure filtered the displacement data and then calculated the higher order derivatives from the displacement data. The second procedure filtered the displacement data with a different cut-off frequency depending upon optimal 0^{th}, 1^{st} and 2^{nd} derivatives. The third procedure involved filtering the displacement data and differentiating for optimal velocity, and then the velocity data was filtered at a different cut-off to obtain optimal acceleration. The last procedure involved differentiation of the displacement data once, filtering the velocity to obtain optimal velocity and then differentiation to obtain optimal acceleration.

The results of the study showed that the first and second procedures produced the lowest rms error and rms error values increased with noise level. The third procedure oversmoothed the data, especially the acceleration, and produced end effects.

Review of these studies suggests that smoothing methods are dependent on noise level, smoothing method, and the derivative calculation. Several of the previous studies used quintic splines instead of a hepatic splines to determine higher order derivatives. The studies emphasize that simulation study is needed to determine an adequate smoothing method.

3.0 SIMULATION STUDY TO EVALUATE SMOOTHING TECHNIQUES

Higher order derivative calculations are very sensitive to errors in measurement [14]. Since this study used markers to track motion at the shoulder, the calculation of jerk required triple differentiation of position. Figure 1 shows a finite difference approximation of jerk plotted versus time for a control. As seen in the figure, the waveform is highly variable with values of jerk ranging in magnitude from 2000 to 12000 cm/sec^3. The large variability in the signal produced a noisy estimate of jerk.

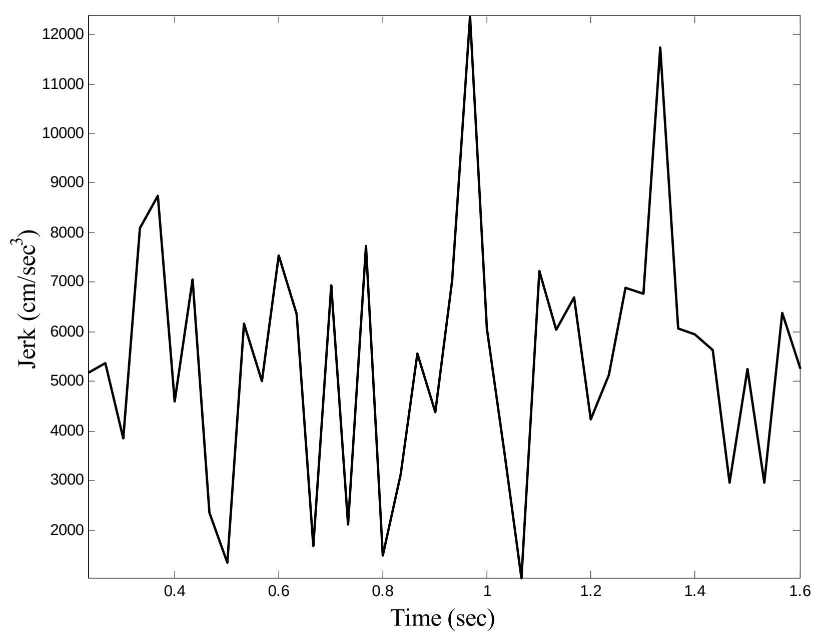

Figure 1: Finite differences approximation of jerk

To obtain usable estimates of the 3rd derivative, smoothing is necessary, but care must be taken to avoid oversmoothing. In order to evaluate smoothing methods for the jerk calculation, a simulation study was performed on a known synthetic trajectory so that the error in the 3rd derivative would be known. The performances of smoothing methods were evaluated by (1) visual comparison of the noise-free approximation of the third derivative with the computed smoothed 3rd derivative and (2) calculation of a signal-to-noise ratio (SNR) and mean squared error (MSE) of the smoothed third derivative signal.

Synthetic shoulder trajectories were created using a hyperbolic tangent equation. Boston et al. showed that this equation could describe hip and knee angles during a repetitive lifting study [8] [12] [13]. This hyperbolic tangent equation was used to describe motion of the shoulder in terms of position as a function of time. The equation is:

$$a(t) = \left[a_i + \frac{(a_f - a_i)}{2} \right] + \left[\frac{(a_f - a_i)}{2} \tanh \frac{(t - t_m)}{t_r} \right]$$

where a_i is the initial (minimum) value of the position data
a_f is the final (maximum) value of the position data
t_m is the time at which the position data amplitude had half the maximum amplitude (midpoint time)
t_r is the time required for the position data amplitude to increase/decrease from 12 to 88 % of the total change.

The hyperbolic tangent equation produces a waveform that closely approximates the y-coordinate of the shoulder position as a function of time. The graphs in Figure 2 show the y-position trajectory for a control subject (top) and a hyperbolic tangent synthetic trajectory (bottom). The synthetic data has the same basic shape of the position trajectory of the subject. The synthetic trajectory was constructed using a midpoint and risetime of 0.8 and 1.2 seconds respectively, which are overall averages of the parameters. Starting and final position along with lift duration (1.9 seconds) were defined from the control subject data shown in the top graph of Figure 2. The 3rd derivative of the synthetic data was computed from first-order central finite differences using four points of the position data. This estimate is called the reference noise-free approximation and is shown in Figure 3.

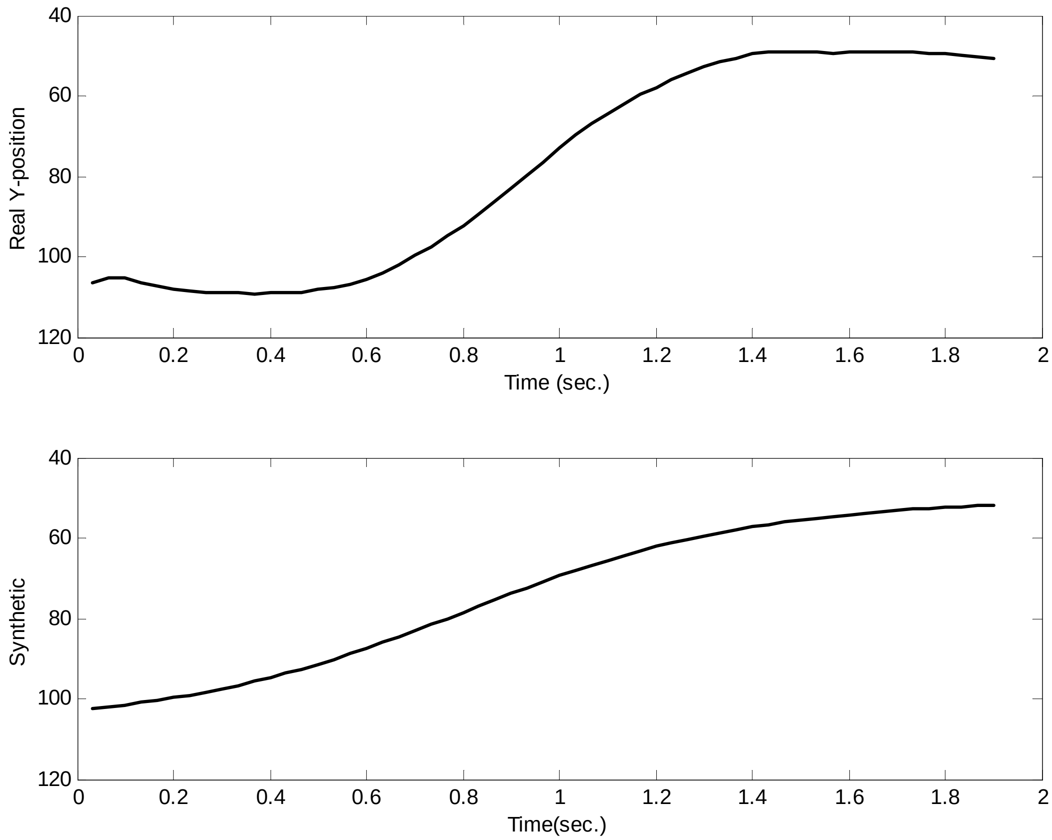

Figure 2: Comparison of a subject's shoulder trajectory as function of time (top) with the synthetic shoulder trajectory approximation as a function of time (bottom)

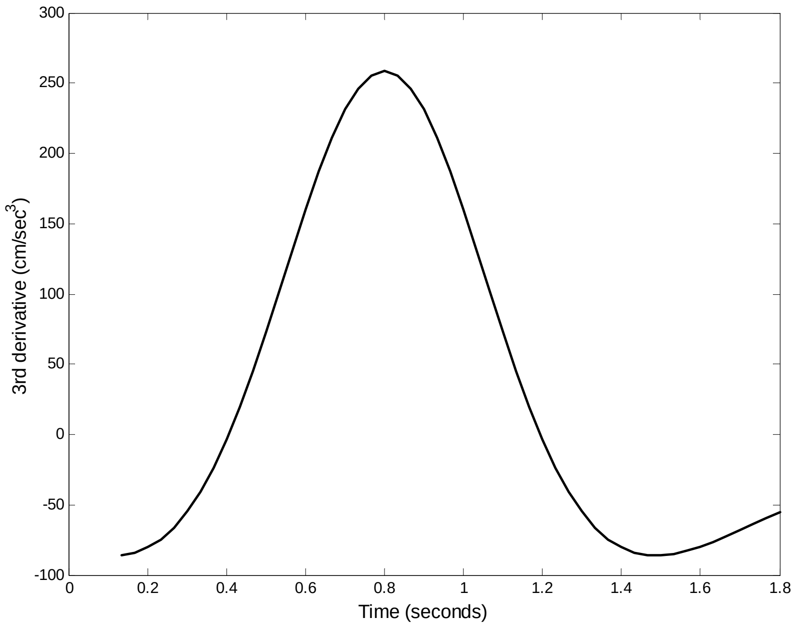

Figure 3: Noise-free approximation of the 3rd derivative

Noise was added to the synthetic trajectory in Fig. 2 to achieve a SNR from 40 dB to 90 dB at 10 dB increments. Noise added to the signal was correlated noise, generated by

$$n_{correlated}(i) = \alpha n_{correlated}(i-1) + n_{white}(i)$$

In the above equation, a random number generator in MatLab produced white noise (n_{white}) and alpha was chosen to equal 0.90. For each of the noise increments added to the trajectory, a SNR was computed as the logarithmic ratio of the rms amplitude of the noise-free signal to the rms amplitude of the noise. The algorithm was run 25 times for each additive noise level with each run producing a different random noise signal. An average of the third derivative SNR and MSE of the 25 runs was computed.

As discussed in Chapter 2, two smoothing methods are commonly reported in the literature to produce suitable estimates of the third derivative: (1) residue analysis and (2) spline fit. Each of these methods was tested on the synthetic data with additive noise, and a 3rd derivative SNR and MSE were calculated.

3.1 RESIDUE ANALYSIS

Low-pass filtering has commonly been employed to eliminate noise from motion data. A low-pass filter rejects higher frequencies that typically represent the noise component in the signal while passing lower frequencies representing the signal. The cut-off frequency of the low-pass filter is determined by finding an appropriate balance between attenuating the noise without distorting the signal [27]. Residue analysis developed by Wells and Winter provides a method for finding an optimal cut-off frequency. The signal is sent through a zero-lag 4th order Butterworth filter at each possible cut-off frequency. The rms amplitude of the difference between the filtered and unfiltered signal (referred to as the residual) is calculated. When the rms of the residual is plotted versus frequency, it appears as a monotonically decreasing function. The optimal cut-off frequency is defined as the point where the noise component is five times the signal component.

The residue analysis was performed on synthetic position data with additive noise and filtered using a zero-lag 4th order Butterworth filter. The rms amplitude of the residues was calculated for cut-off frequencies from 1-14 Hz. Figure 4 shows the rms residuals plotted versus cut-off frequency for the synthetic data at additive noise ranges of 40-90 dB. The optimal cut-off frequency is 1.3 Hz, since at this frequency the noise component is five times the signal component.

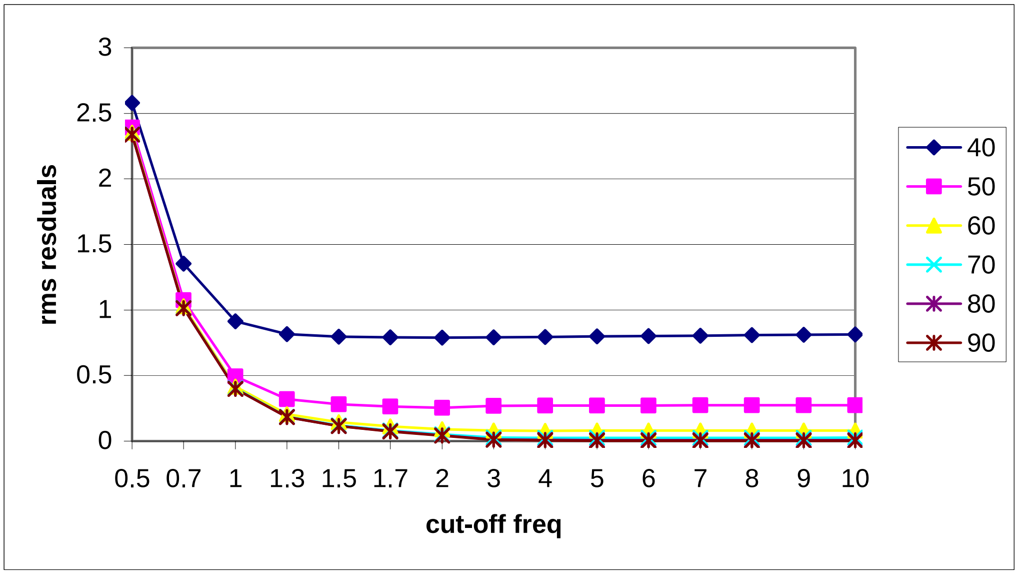

Figure 4: RMS residuals versus cut-off frequency. Diamonds indicate 40 dB noisy trajectory, blocks indicate 50 dB noisy trajectory, triangles indicate 60 dB noisy trajectory, the x's indicate 70 dB noisy trajectory, stars indicate 80 dB noisy trajectory and asterisk indicate 90dB noisy trajectory

Before differentiation, the noisy displacement data were filtered using the Butterworth filter at the optimal cut-off frequency. All derivatives were calculated using finite differences. The velocity was calculated using a forward difference, acceleration as a three-point central difference using displacement data and the 3rd derivative as a forward differences using acceleration. This finite difference equation combination was determined by trial-and-error to provide a better estimate of the third derivative than other combinations of finite difference equations. The 3rd derivative was then filtered with the Butterworth filter at the optimal cut-off frequency. The filter was only applied to the displacement and 3rd derivative. Applying the filter to 1st and/or 2nd derivative and then calculating and filtering the 3rd derivative increased variability.

3.2 SPLINE

The synthetic shoulder trajectory was fit to a natural hepatic B spline generated by Woltring's generalized cross validation algorithm (GCVSPL) in MatLab. A hepatic spline was chosen to satisfy the constraint that half the spline order should be higher than the highest derivative sought [14]. The input trajectory is used to define the values of the spline at the knots. For this study, time at each position was used as the knots of the spline. The spline then essentially interpolates for the position by fitting a piecewise continuous polynomial between two knots (two time points) so that the spline fit is now defined for points between the knots of the original trajectory.

The Woltring spline algorithm requires a smoothing parameter 'μ'. The smoothing parameter is specific to the data being fit to the spline and must be determined through application to a test signal. Values of μ from 10^{-8} to 10^{-5} were tested on the synthetic trajectory with additive noise. Derivatives were calculated by differentiating the spline coefficients using MatLab program SPLDER.

The parameter μ was determined by comparing the 3rd derivative SNR for various μ values. The average 3rd derivative SNR plotted versus the log of μ is showed in Figure 5. The 3rd derivative SNR increases as the values of μ become larger for noise ranges of 40-60 dB. The 3rd

derivative SNR of 70-90 dB shows an increase of the SNR to a peak at a smoothing parameter of $10^{-6.5}$, and the 3rd derivative SNR deceases for larger values of μ. The decrease in the 3rd derivative SNRs after the peak indicates that these values of μ oversmoothed the data.

The noisy synthetic data was compared to the real data to estimate the level of noise. The comparison showed the real data had a noise level that was approximately 70 dB. Therefore, $\mu = 3 \times 10^{-7}$ ($\approx 10^{-6.5}$) was used in the algorithm of the hepatic spline fit.

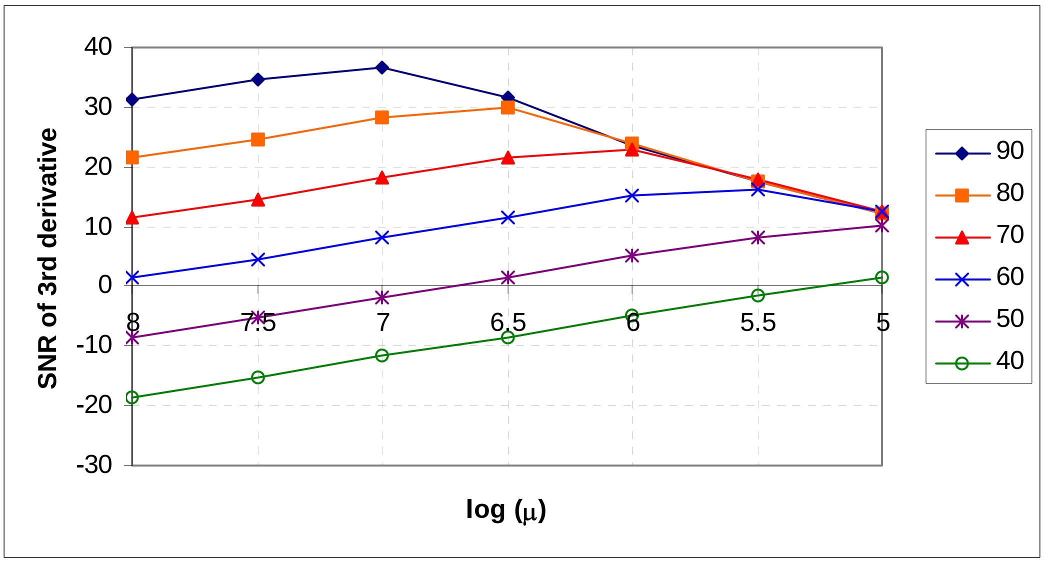

Figure 5: 3rd derivative SNR plotted versus smoothing parameter of the hepatic spline fit. The top diamond line is additive noise of 90 dB, square line is 80 dB, triangle line is 70dB, 'x' line is 60dB, asterisk line is 50 dB, and the bottom circle line is 40 dB.

3.3 COMPARISON OF SMOOTHING METHODS

Both of the smoothing methods provided suitable estimates of the 3rd derivative when compared to the direct estimate obtained from finite differences. Figure 6 shows the direct estimate of the 3rd derivative for the synthetic trajectory with 70 dB of additive noise. The 3rd derivative estimate of the synthetic trajectory is similar to the 3rd derivative estimate obtained for subject data in Fig. 1. Both figures demonstrate large variability and are too noisy to be useful estimates of the 3rd derivative. Figures 7-12 show the 3rd derivative with smoothing plotted as a function of time for each of the noise levels. Even though both smoothing methods reduced noise, the hepatic spline was chosen as the smoothing method for the subject data since the spline produced a higher SNR and lower MSE than the residual analysis as shown in Table 1. The residual analysis underestimated the noise-free approximation for all additive noise levels. Both methods demonstrated end effects, which resulted in overestimation of the noise-free approximation at the boundaries of the waveform. End effects increased as the SNR decreased. At noise levels of 50 dB and lower, the hepatic spline performed poorly. The residual analysis provided a better estimation of the noise-free approximation at SNRs of 40-50 dB.

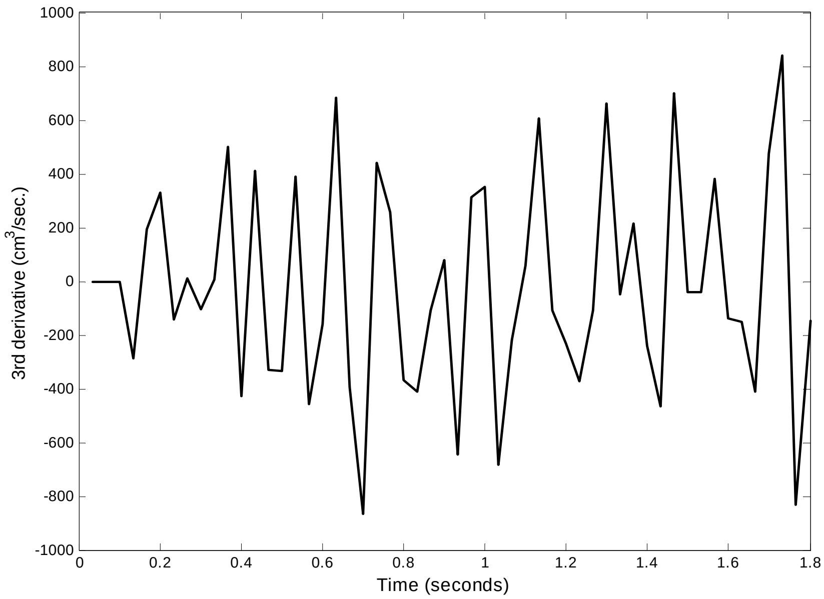

Figure 6: Direct estimate of the 3rd derivative of the synthetic trajectory with 70 dB of additive noise

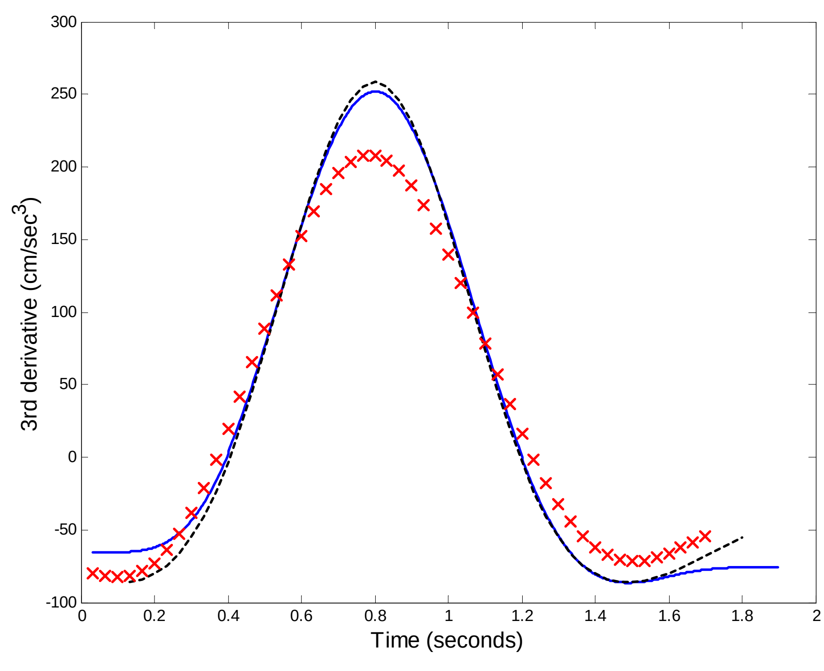

Figure 7: Comparison of smoothing methods at 90dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

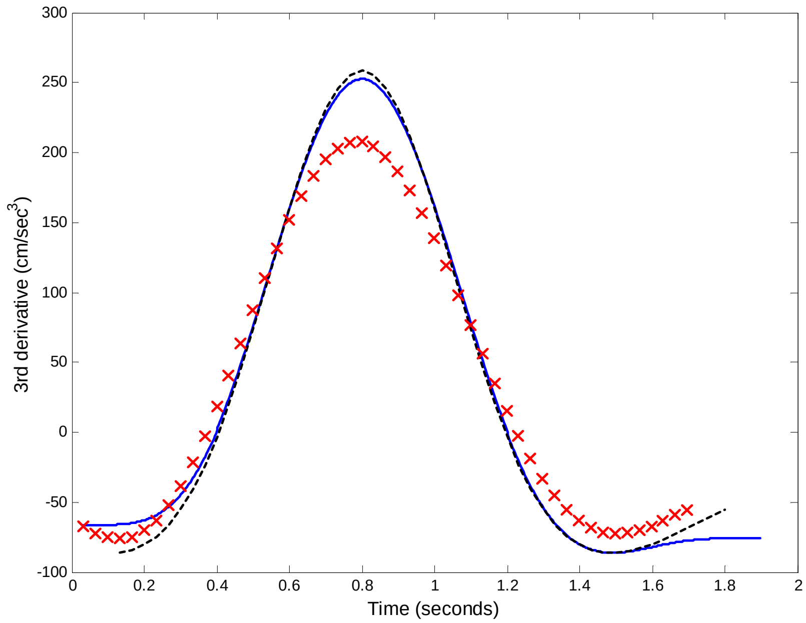

Figure 8: Comparison of smoothing methods at 80dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

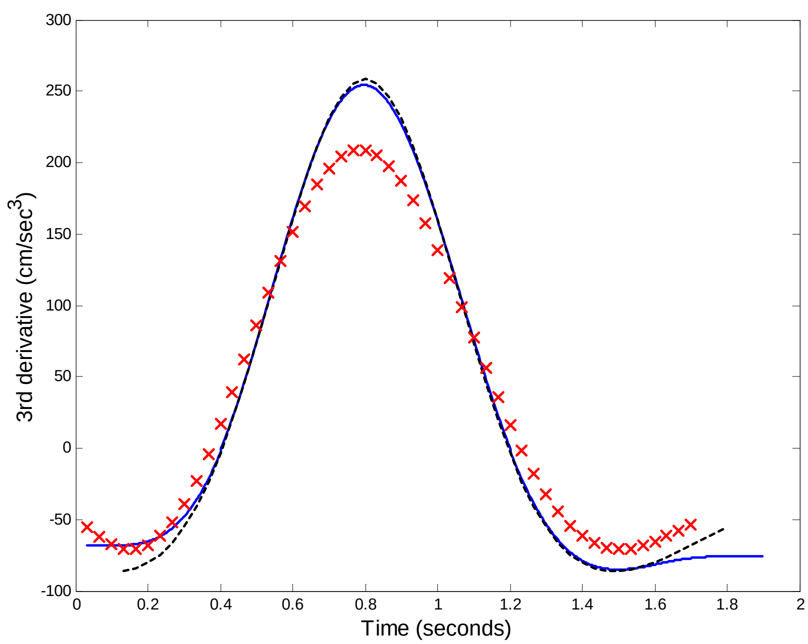

Figure 9: Comparison of smoothing methods at 70dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

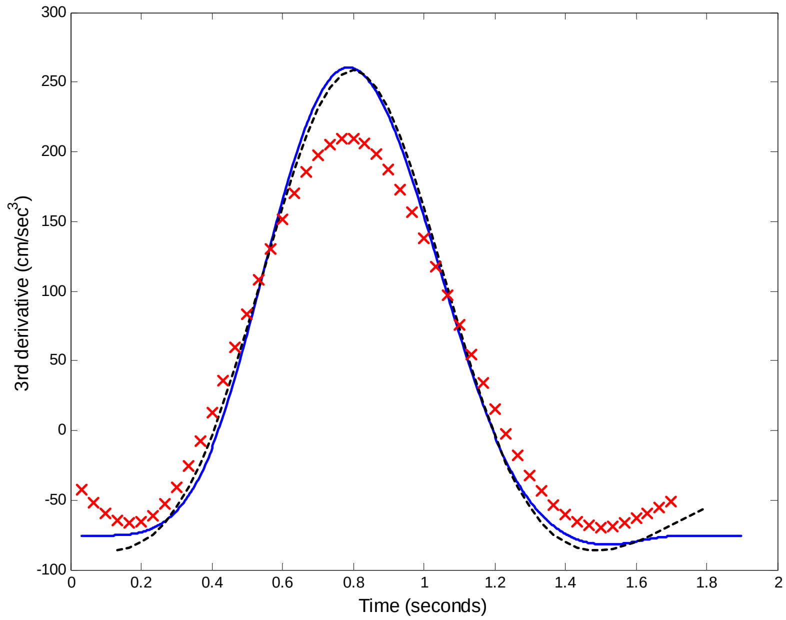

Figure 10: Comparison of smoothing methods at 60dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

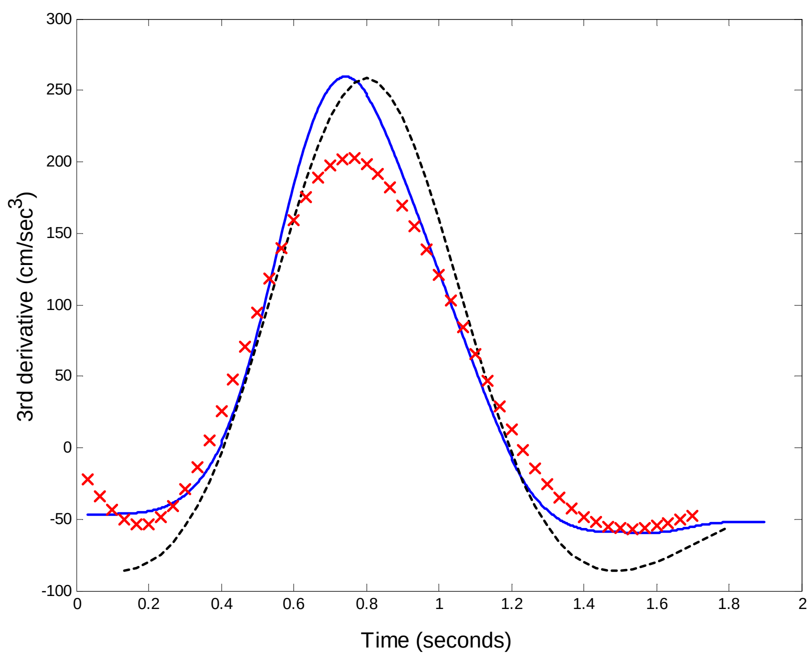

Figure 11: Comparison of smoothing methods at 50 dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

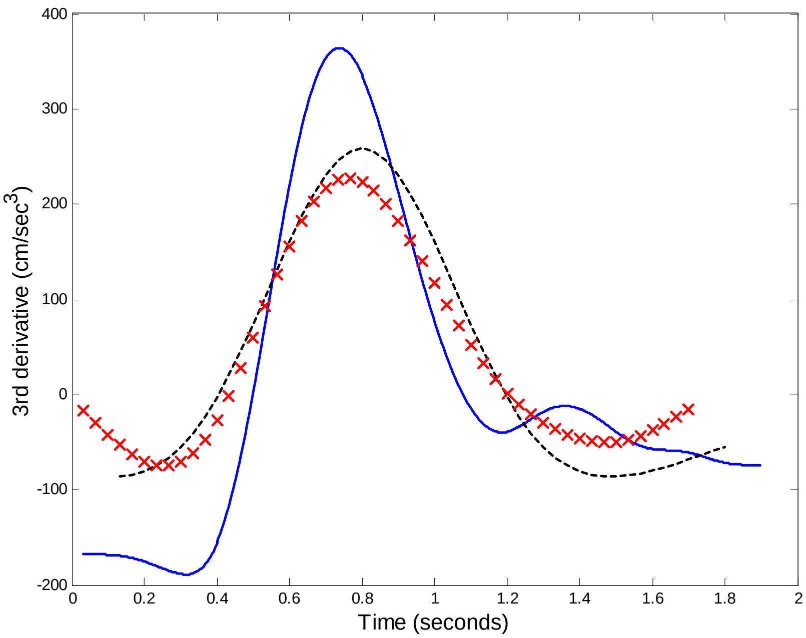

Figure 12: Comparison of smoothing methods at 40 dB of additive noise. Noise-free approximation is the dashed line, hepatic spline is the solid line and the residual analysis is the 'x' line.

Table 1: 3rd derivative SNR and MSE for each smoothing method

	3rd derivative SNR			3rd derivative MSE		
Additive noise (dB)	Direct estimate	Hepatic spline	Residual Analysis	Direct estimate	Hepatic spline	Residual Analysis
40	-35.92	5.24	6.23	1.94×10^8	17088	10721
50	-25.92	14.42	6.95	1.94×10^7	2038	10216
60	-15.95	20.22	7.35	1.96×10^6	470	9269
70	-6.18	21.79	7.62	2.06×10^5	293	8697
80	1.96	22.08	7.83	3.06×10^4	269	8297
90	5.53	22.14	7.97	1.32×10^4	264	8031

Extrapolation of the data by sampling several seconds before and after the lift starts can reduce the end effects. Hreljac started sampling data several seconds before the execution of the heel contact during his running task [28]. Vint and Hinrich extrapolated as many as 20 points to the ends of the data, while Giakas and Baltzopoulos used 10 points padded with zeros [32] [15] [34]. The data in this study were sampled to obtain a padding of 8 points at the beginning and end of the data. This padding was not sufficient to eliminate all end effects.

As discussed in Chapter 2, several investigators found the residuals analysis provided better estimates of the 3rd derivative than the spline fit. The discrepancy between results of this project and previous studies could be noise level. The residual analysis performed better than the spline fit for SNRs of less than 50 dB. For this project, the noise level in the subject data was estimated to be 70 dB by comparing the synthetic trajectory to subject data. It is possible that in the previous studies, the data had SNR less than 50dB and thus the residual analysis was the better smoothing method.

Another difference between this project compared to previous studies is that a hepatic spline was used instead of a quintic spline. The quintic spline imposes a zero boundary condition on the 3rd derivative. As stated in Chapter 2, if the zero boundary condition is violated, error and variability will occur throughout the data and not just at the boundaries, leading to poorer performance of the spline method.

4.0 EXPERIMENTAL METHODS

A functional capacity assessment involving a repetitive lifting task has been designed to quantify motion differences between CLBP patients and pain-free controls [8]. During the lifting task, subjects repeatedly lift a resistant load and reflective markers track joint motion. From the motion data, jerk and several other performance parameters were calculated to describe lifting patterns. The lifting task was applied to CLBP patients before and after treatment, since each of the CLBP patients was required to complete a rehabilitation program after the first functional capacity assessment. Statistical methods determined differences in the parameters due to groups, treatment and changes over time during the task.

The data used for this project was collected as part of a clinical study that described lifting movements using several different performance measures [13]. The clinical study was conducted at the University of Pittsburgh Pain Evaluation and Treatment Institute and was constructed to determine the impact of CLBP on daily activities. Subjects answered questionnaires about their pain, performed tests assessing functional mobility and participated in a repetitive lifting task.

4.1 SUBJECTS

Seventy-seven subjects, 54 pain-free control and 23 CLBP patients, participated in the repetitive lifting study. The age of the subjects ranged from 36-63 years and both groups were approximately matched for gender. All subjects gave written informed consent as approved by the University of Pittsburgh Biomedical Institutional Review Board before the start of the lifting task. In the CLBP patient group, all of the 23 patients had a history of prolonged back pain with mean pain duration of 3.2 years (standard deviation = 4.4 yrs.) Fifty-three percent of these patients had myofascial back pain diagnosis, 34 % had a diagnosis of mechanical low back pain and the remaining 13% had various other diagnoses. Additionally, all but one CLBP patient reported experiencing lower back pain on a daily basis and 29% reported at least one pain-related surgery. The pain-free control group was composed of adult volunteers with no medical history or current complaint of back pain. College students and competitive athletes were excluded from the control group.

A physician and physical therapist evaluated all subjects to determine his or her ability to participate in the repetitive lifting task. The physical therapist completed a brief assessment of each subject's functional mobility and upper extremity range of motion. No subject was excluded based on this functional mobility examination.

4.2 PROTOCOL

The repetitive lifting task required each subject to lift a handle attached to a resistant load located 13 inches from the ground to waist height. The BTE Work Simulator (Baltimore Therapeutic Equipment Company, Baltimore, MD, USA) provided the resistant force for the up-phase of the lift. Subjects performed lifts for a total of twenty minutes with a fifteen second rest interval between each of the lifts, during which the subject returned to the standing position. Four hemispheric infra-red reflective markers, placed on the left side of the body, tracked joint motion throughout the lifting task. Markers placement was on the ankle, apex of the patella, greater trochanter of the femur, and acromion of the shoulder. A schematic of the lifting task is shown in Figure 13.

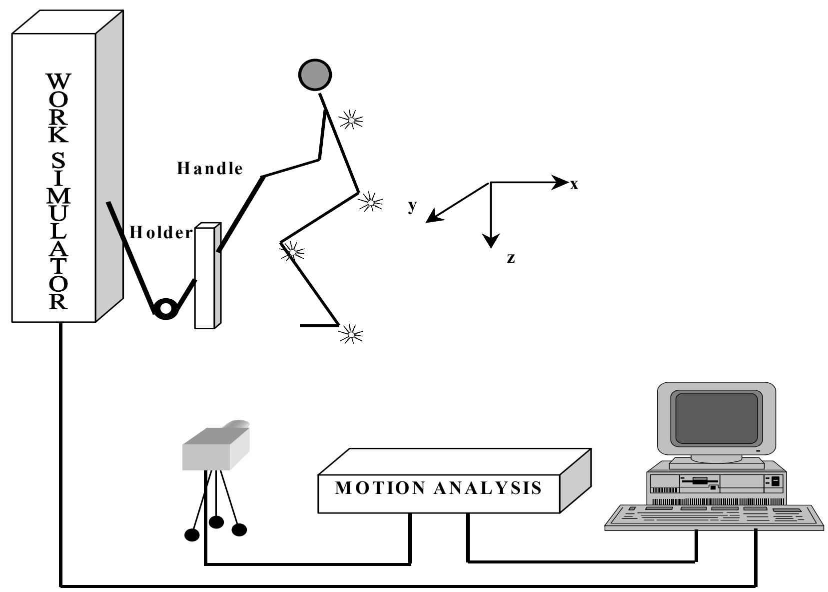

Figure 13: Schematic of lab set-up

The resistant force applied during the lifting task was equal to 40% of each subject's maximum voluntary static strength. Maximum voluntary static strength was measured with a force gauge (Chatillon Muscle Strength Dynamometer, Sammons Preston, Bolingbrook, IL, USA) attached to a platform. Subjects were instructed to assume a bilateral symmetrical leg lift position with the forearm in supination and the handle of the force gauge adjusted to knee height. The subject was then instructed to pull steadily on the force gauge for approximately four seconds. This process was repeated three times with a fifteen second rest period between each attempt, during which the subject was instructed to return to a standing position. Forty percent of the mean of the three trials was used as the resistant force.

Before the start of the repetitive lifting task, each subject was given the opportunity to practice the lift without resistant load applied to the handle in order to become familiar with the task. Once the subject was comfortable, the resistant load was applied and the repetitive lifting task began. Throughout the experiment the subject was given no verbal or visual feedback concerning performance. The task was terminated: (1) if the subject felt physically unable to continue, (2) if the experimenter stopped the task due to unsafe body biomechanics or (3) the time limit was reached. Once the task was terminated, CLBP patients were asked to rate their pain using the pain severity scale from the multidimensional pain inventory [35].

Each of the CLBP patients, after completion of the lifting task, was enrolled in an intensive rehabilitation program. Patients attended the Pain Evaluation and Treatment Institute daily for 8 hours for a 3 ½ week period. The rehabilitation program focused on education about body mechanics, physical exercises to increase endurance, and flexibility. Treatment included a combination of group and individual physical, occupational, and psychological therapies [36]. During treatment, subjects did not have any training on the BTE Work Simulator. Once the rehabilitation program was completed, the CLBP patients were retested using the same lifting protocol as before. The two separate testing sessions were labeled as pre-treatment assessment and post-treatment assessment of CLBP patients.

Twenty control subjects repeated the lifting protocol after 3½ weeks to determine reliability of measurements. For the control subjects, testing sessions were labeled as baseline assessment and repeat assessment.

4.3 INSTRUMENTATION

The BTE work simulator provided the resistance for each subject to work against during the repetitive lifting task. The work simulator is a computerized device that maintains constant force on the handle during the lift. The force transmission occurs at the handle by a rope connected through a pulley system. The starting height, waist level (ending height) and force were programmed into the work simulator before beginning the task. A series of tones instructed the subject when to lift and when to lower the handle. A high tone indicated that the rest period was over and the subject was to perform a lift. A low tone indicated that the handle had reached waist height and the subject could return the handle to the holder. A second, lower tone indicated that the subject had placed the handle in the holder, initiating the rest period. The BTE software (Baltimore Therapeutic Equipment Company, Baltimore, MD, USA) allowed the subject to perform the lifts at his or her own pace. The work simulator recorded the force and handle velocity at a sampling rate of 50 samples per second.

Motion Analysis Model 110 Video Processor using Expert Vision Software (Motion Analysis Corporation, Santa Rosa, CA, USA) and an NEC TI-23A CCD camera with LED ring-light tracked the retro-reflecting markers attached to the subject. The motion analysis system tracked the markers at 30 frames per second during the up-phase of the lift by detecting the marker boundaries.

4.4 JERK

Jerk was calculated for the shoulder trajectory obtained from the shoulder marker displacement. The shoulder trajectory was chosen because shoulder motion, depending on motion at both the hips and knees, integrates motion of both upper and lower body. Using the shoulder to calculate jerk is consistent with the study of Young and Marteniuk. A typical shoulder position trajectory as a function of lift time is shown in Fig. 14. The lift starts with the shoulder at position indicated at the lower left corner of the graph. At this position, the subject was beginning the lift by grasping the handle. The lift ends when the subject was standing at full extension with ending shoulder position indicated at upper right corner of the graph. The duration of this lift was 1.867 seconds.

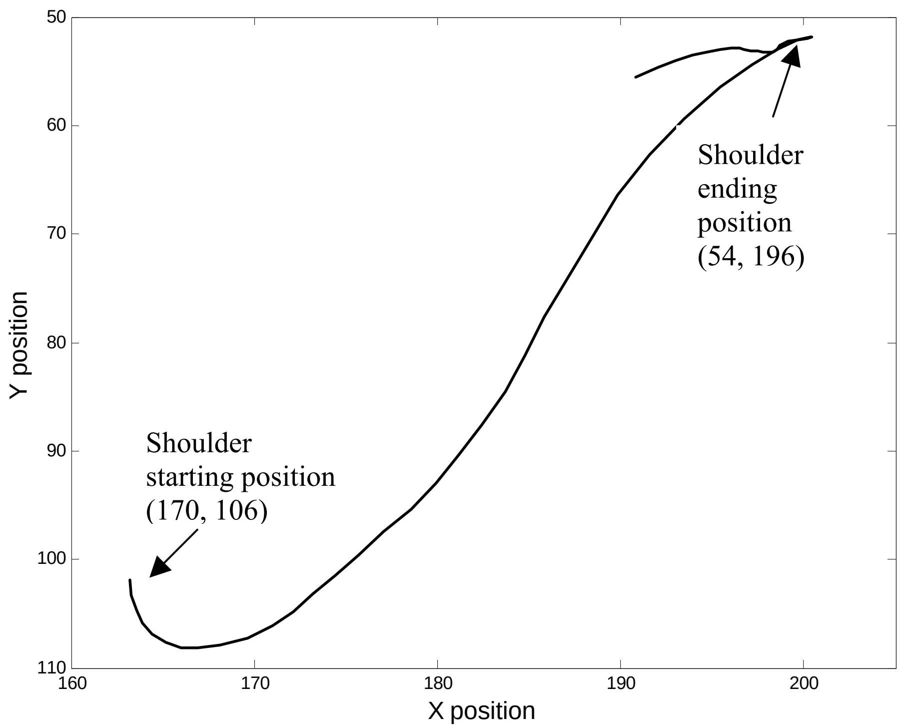

Figure 14: X-position versus Y-position for a control's shoulder trajectory

Jerk, J(t), is defined as the magnitude of the third derivative of the x and y coordinate position and is shown in the following equation:

$$J(t)=\sqrt{\left(\frac{d^3x}{dt^3}\right)^2+\left(\frac{d^3y}{dt^3}\right)^2}$$

Jerk was calculated as function of lift time for each lift of each subject. The root-mean-squared amplitude of jerk was then calculated over each lift for each subject. Rms jerk was used to reduce variability, since rms function averaged jerk over the entire waveform.

Woltring's generalized cross-validation B spline algorithm (GCVSPL) in MatLab (MathWorks Incorporated, Natick, Massachusetts, USA) was used to fit a hepatic spline to the shoulder trajectory of each subject. Woltring originally wrote the program to run in Fortran (Fortran Company, Tucson, Arizona, USA) and Tian, Carta and Reina adapted the code to MatLab (www.isbweb.org downloaded 1/27/2003). The program assumes additive, uncorrelated noise, and essentially smooth, underlying functions. In the program, the amount of smoothing can be explicitly given, based on the Generalized Cross-Validation and Means-Squared Prediction Error Criteria of Craven and Wahba [30] or based on the effective number of degrees of freedom [37]. Each of these smoothing options was examined on the data. For this project, the option of providing a smoothing parameter was chosen, since this option was the most effective in reducing noise. The amount of smoothing was given as 3×10^{-7} and was determined through a simulation study described in Chapter 4. Derivatives were found by differentiating the spline coefficients using the MatLab program SPLDER written by Woltring, Tian, Carta and Reina (www.isbweb.org downloaded 1/27/2003).

4.5 DATA ANALYSIS

The data processing procedures began by tracking markers throughout the task, producing a centroid file that contains the coordinates of each marker for each frame. A MatLab program matched and assigned each of the coordinates of each frame to a specific marker. From this program, position of the shoulder as a function of time during the lift was determined. The position trajectory was fit to a hepatic spline and jerk was calculated as a function of time. The fit of the hepatic spline to the shoulder trajectory and calculation of the derivatives are illustrated in Fig.15 for a control. The top left graph is the y-position of the trajectory plotted versus time and the top right graph is the x-position of the trajectory plotted versus time. Directly below each of these graphs are the spline fit to the position data for the x- and y-position versus time respectively. In comparing the graphs, the experimental data contains a small amount of noise that the spline fit effectively eliminates without distorting the shape of the trajectories. Also, comparison of the graphs demonstrates the interpolation function of the spline fit that increases data points. The third row shows the 1st and 2nd derivatives of the spline fit to the experimental data. The left graph is the 1st derivative of the x-position and y-position versus time and the right graph is the 2nd derivative of the x and y position versus time. In all the graphs, the x-position derivative is a solid line and the y-position derivative is a dashed line. The last row of graphs is the plot of the 3rd derivative (left) and jerk plotted as a function of time (right). As seen from the figure, the hepatic spline smoothing provided a useable estimate of the third derivative.

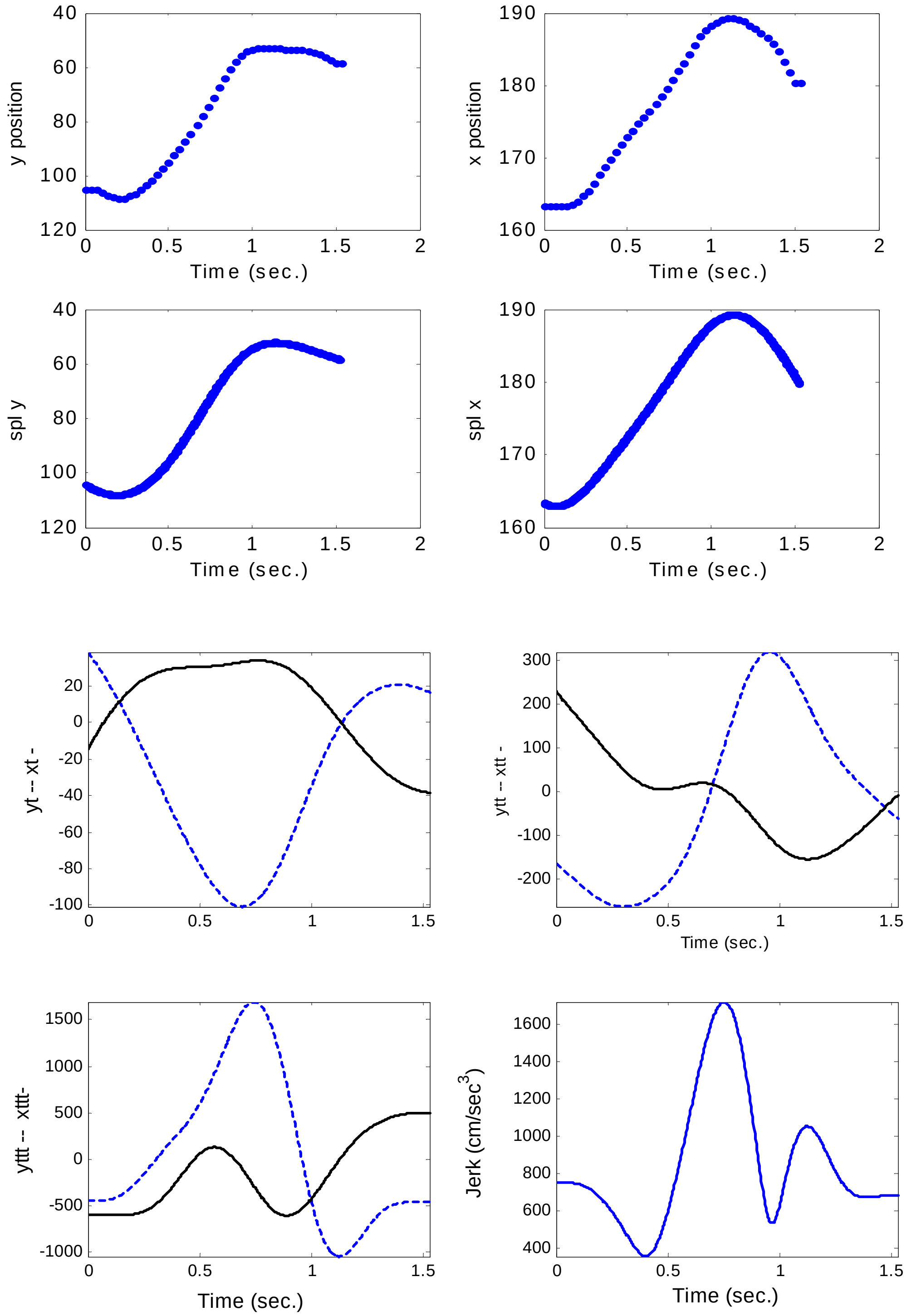

Figure 15: Data Analysis. Top graph shows the shoulder x- and y-position trajectories as a function of time, and the two graphs below the top graphs show hepatic spline fit to the shoulder trajectories. The middle graphs are the 1st and 2nd derivatives as a function of time and last two graphs are 3rd derivative and jerk as a function of time

4.6 BODY ANGLES

A two-dimensional three-segment biomechanical model was constructed from the motion of the four joint markers. The three segments were defined as the shank, thigh and trunk. From the model, joint angles were defined from the segment angles. The knee angle was defined by the angle between the shank and thigh segments; and the hip angle was defined by the angle between the thigh and trunk segments. Full extension was defined as zero degrees.

4.7 PERFORMANCE MEASURES

The performance measures analyzed were weight lifted, number of lifts completed, starting posture, midpoint difference, risetime difference and lift duration. Weight lifted and lifts completed are performance outcome measures, since these measures describe performance for the entire lifting task. Lifts completed were the number of lifts the subject performs during the task.

Lift duration, starting posture index, midpoint difference and risetime difference are performance style measures which describe basic body biomechanics used by the subject for each lift. Lift duration is the time during which the BTE Work Simulator applied resistance to the handle. An index of starting posture was derived as starting hip angle minus starting knee angle divided by the starting hip angle, creating an index that ranged from -1.0 to 1.0. Values approaching -1.0 indicate a starting posture characterized by a squat lift in which the back is kept vertical and the hip and knees are flexed at the start of the lift. Values near 1.0 indicate a torso style lift in which the back and hips are flexed and the knees are kept straight at the start of the lift. Values around zero indicate a freestyle lift in which both the back, hip, and knees are flexed [13].

4.8 STATISTICS

RMS jerk, calculated using a hepatic spline fit to the shoulder trajectory, was used as a parameter to describe lifting patterns. Statistical analyses were performed to assess whether this particular method of calculating jerk can detect differences due to group, treatment, and changes over time. Three different comparisons of rms jerk were constructed and tested: (1) rms jerk between controls at baseline and patients pre-treatment assessment, (2) rms jerk between patients pre-treatment and patients post-treatment assessment, (3) rms jerk between controls baseline and patients post-treatment assessment. A mixed model repeat measures multivariate analysis of variance (MANOVA) tested each of the comparisons. A p-value of less than 0.05 was considered to be statistically significant for all analyses.

The performance measures have been evaluated previously. However since the subjects included in this study were a subset of the subjects used in the previous study, the statistics for these measures were processed for this set of subjects. Each of the performance measures was tested for statistical differences due to group, treatment and changes over time.

To assess temporal changes of the parameters, lifts were separated into phases. For each subject, lifts were grouped as early, middle, or late phases, based on the individual subject's number of repetitions [13]. Averages of rms jerk and the performance style measures were computed over 5 lifts in the early phase of the task, over 5 lifts in the middle phase of the task, and over 5 lifts at the late phase of the task for each subject. The phase separation is demonstrated in Fig. 16, which is a plot of rms jerk as a function of lift number with lines connecting data points. The early phase is the average of rms jerk for lifts 1-5, middle phase is the average of rms jerk for lifts 33-37 and late phase is the average of rms jerk for lifts 65-70. The phase separation reduces variability within the parameters by averaging adjacent lifts and normalizing differences due to the number of lifts completed.

The number of lifts a subject completed was dependent on how fast the subject lifted, since the rest period is initiated once the handle is placed into the cradle. A subject that lifted at a high speed completed more lifts within the 20-minute testing than a subject that lifted at a slow speed. Thus, the phases created a method of comparing performance over varying number of lifts. Using the phases in a repeated measures analysis examined not only the group differences but also the changes over task phase and group-by-phase interactions. The multivariate results of the repeated measures ANOVA were used, since variance cannot be assumed to be equal.

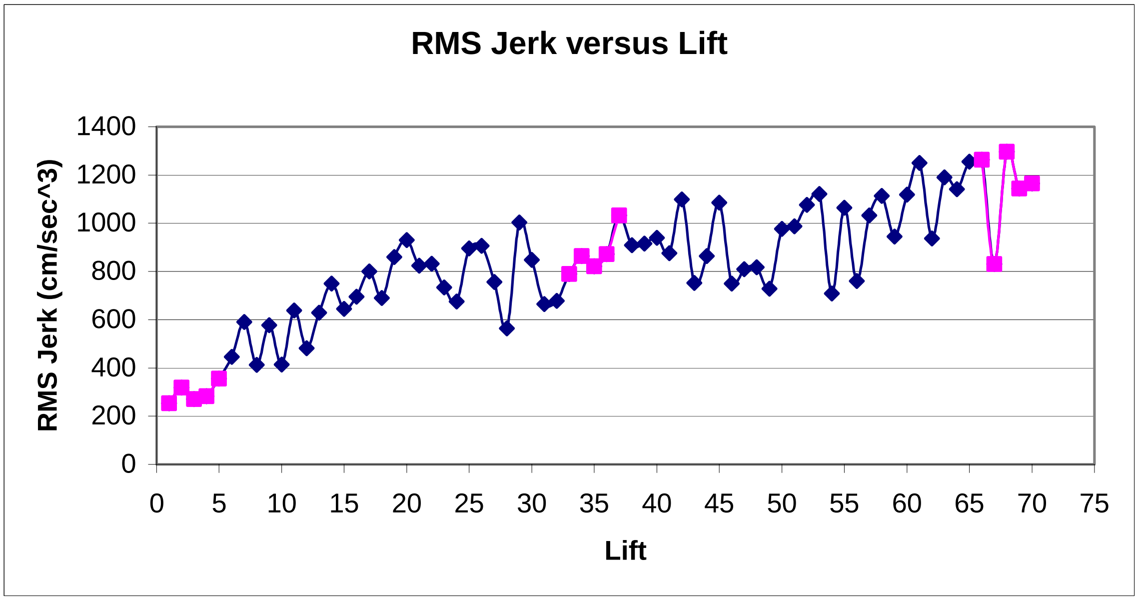

Figure 16: RMS jerk plotted versus lift number. The lighter squares indicate the phase separation for the parameter averages.

In order to compare rms jerk between treatment assessments, reliability of rms jerk must be assessed. Reliability was determined by comparing baseline with repeat assessment of control subjects with repeated measures MANOVA and calculation of an intra-class correlation. RMS jerk and performance style measures were calculated for each lift of each assessment. The parameter values for each phase were calculated and tested for significant differences between assessments and changes over task phase. This design and corresponding data analyses represented a two-within component of assessment (baseline vs. repeat) and phase (early, middle and late) completely crossed design.

The intra-class correlation was calculated for jerk at each phase of the task and for an overall mean of the three phases. The correlation compared baseline with repeat assessment of the control. The time between assessments was 3 ½ weeks, which is the same as the time period between treatment assessments of CLBP patients. A p-value was calculated to determine whether the intra-class correlation was significantly reliable. A separate p-value was calculated at each phase to determine if any significant difference between the two assessments occurs, indicating a bias. An intra-class correlation was not performed on the performance style measures, since all of the performance style measures were previously shown to have good reliability [8].

The first comparison determined group differences and changes over task phase of rms jerk and the performance style measures by comparing pre-treatment CLBP patients with baseline controls. Parameter values at each phase were used as the repeat measure in a MANOVA [38]. The basic experimental design and corresponding data analyses represented a one-between (control vs. pain subjects) and one-within (phase - early, middle, and late), completely crossed design.

To control for the possibility that lift duration can confound the interpretation of differences in rms jerk, a Pearson linear correlation followed by an analysis of covariance (ANCOVA) was performed on the lift duration and rms jerk. The ANCOVA was designed using an average of early, middle and late phases of lift duration as a covariate and rms jerk as the dependent measure. This design tested for significant differences between groups and changes over time after adjusting for lift duration.

A Pearson correlation was performed between jerk and the pain severity rating obtained at the end of the task for CLBP patients. The correlation was performed to determine whether reported pain severity was related to patients' performance of jerk during the repetitive lifting task.

The effect of gender on the parameters was tested to eliminate any possible bias of group differences caused by gender. A repeated measures MANOVA was performed for all parameters collapsing across groups. The MANOVA tested for parameter differences between gender and changes over time. The basic experimental design and corresponding data analyses represented a one-between (male vs. female subjects) and one-within (phase - early, middle, and late) completely crossed design.

The second comparison determined the impact of treatment by comparing CLBP patients pre-treatment with post-treatment assessment. For each assessment of CLBP patients, parameters were calculated for each lift. The parameter values at each phase were calculated and tested for significant differences between assessments and changes over task phase. This design and corresponding data analyses represented two-within of assessment (pre-treatment vs. post-treatment) and phase (phase-early, middle and late) completely crossed design.

The final comparison determined whether after rehabilitation, patients would perform lifts similar to controls. This comparison compared post-treatment patients with controls baseline assessment. The experimental design and corresponding data analyses represented a one-between (control vs. patient subjects) and one-within (phase - early, middle, and late) completely crossed design.

Effect size index was calculated for rms jerk and the performance style measures. Effect size was utilized to determine how large a difference between group means could be detected by the study, after adjusting for the magnitude of the variable. The larger the effect size of a dependent variable, the more likely a statistical significance will be attained and the greater the statistical power [39]. Effect size is calculated as the difference in the means divided by a pooled standard deviation. An effect size range of 0-0.32 is small, 0.33-0.55 is medium and 0.56-1.2 is large [39]. Effect size was determined for each parameter for comparison between baseline controls and pre-treatment patients, comparison between pre-treatment patients and post-treatment, comparison between patients post-treatment and controls baseline, and for comparison of baseline controls and repeat controls.

5.0 RESULTS

Statistical differences of group, treatment and changes over time of jerk and the performance style measures are described in this chapter. Subject demographics, reliability of jerk and performance outcome measure differences between groups are also described.

5.1 SUBJECT DEMOGRAPHICS

Subjects were tested for significant differences in gender, age, weight and height. No significant difference was found between the groups for any of the demographics. Table 2 lists the mean and standard deviation of each of the demographics. A MANOVA was constructed to determine whether gender had any influence on lifting style. Significant differences in gender were only seen in the weight lifted. Males lifted more weight than female subjects ($p = 0.001$) as shown in Table 3. There existed no group-by-gender interaction for weight lifted. RMS jerk, starting posture, lift duration, midpoint and risetime differences and number of lifts completed all demonstrated no statistical significance for gender, gender-by-phase and group-by-phase-by-gender interaction.

Table 2: Average values (standard deviation) of subject demographics

Subject Demographics				
	Male		Female	
	Control	Patients	Control	Patients
Sample size	26	12	28	12
Age (yrs.)				
Mean	35.1	37.9	35.3	37.6
SD	(8.1)	(9.4)	(13.2)	(11.5)
Weight (kg.)				
Mean	82.4	86.2	64.1	70.2
SD	(13.3)	(16.7)	(10.4)	(12.4)
Height (cm.)				
Mean	178.6	175.8	165	165.8
SD	(11.7)	(12.1)	(7.1)	(7.6)

Table 3: Average values (standard deviation) of performance outcome measures comparing controls at baseline and patients at pre-treatment assessment

Performance Outcome Measure							
	Group		Gender		P-value		
	Control	CLBP	Male	Female	Group	Gender	Gr by Ge
Weight lifted (kg) (SD)	28.4 (14.7)	16.9 (12.6)	33.8 (15.5)	16.4 (7.8)	0.0001	0.0001	0.194
Number of lifts (SD)	68.4 (8.98)	41.1 (16.4)	60.21 (16.8)	60.23 (17.6)	0.0001	0.625	0.65

5.2 RELIABILITY

A subset of controls repeated the lifting task, resulting in two assessment times: baseline and repeat assessment. No statistical difference was found for rms jerk when comparing baseline and repeat assessment of controls ($p = 0.089$). A difference across task phases was significant ($p < 0.0001$). Controls at both the baseline and repeat assessment increased rms jerk as the task progressed. No assessment-by-phase interaction occurred between baseline testing and repeat assessment ($p = 0.372$). The mean rms jerk values for each assessment of controls along with the standard deviation, effect size and p-values are listed in Table 4.

Repeat assessment of control subjects showed no assessment effect for weight lifted and number of lifts completed as shown in Table 5. Controls decreased lift duration at the repeat assessment from the baseline assessment ($p < 0.0001$) and continued to lift faster as the task progressed ($p < 0.0001$). An assessment-by-phase interaction existed ($p = 0.003$), as shown in Figure 17. Controls lifted faster at each phase during the repeat assessment. Repeat assessment showed no differences for starting posture, midpoint and risetime differences for assessment, phase and assessment-by-phase.

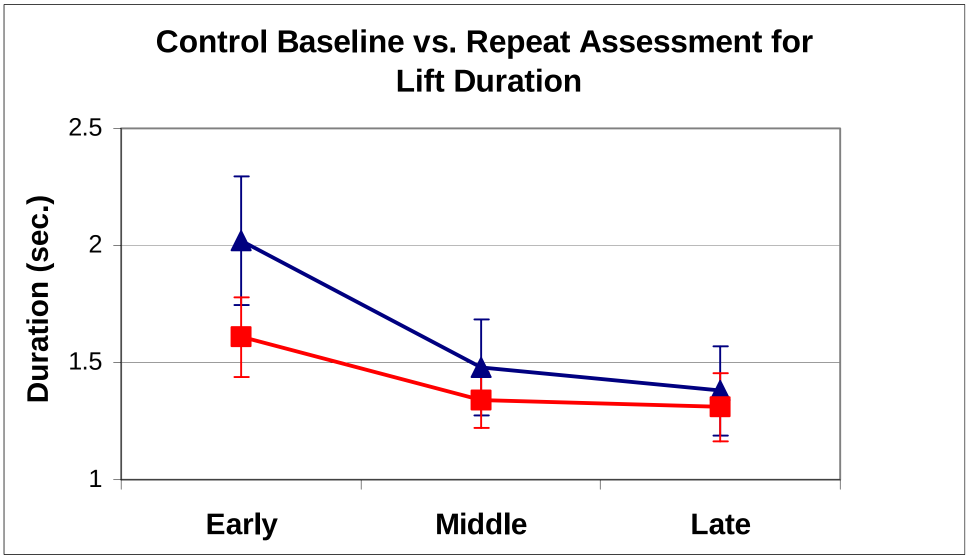

Figure 17: Plot of lift duration at each phase comparing assessment of controls. ▲: Controls at baseline and ■: Controls at repeat assessment

An intraclass correlation coefficient was utilized to determine the stability of jerk with repeat testing. The reliability analysis of rms jerk was calculated for early, middle and late values and for the mean of early, middle and late values. All analyses of rms jerk had moderate reliability except the late phase of jerk, which produced an intraclass R of 0.281 ($p = 0.11$). Mean rms jerk had a significant intraclass R of 0.4148 at a p-value of 0.022. Early and middle phase rms jerk produced a significant intraclass R-value of 0.484 and 0.442 respectively. The early phase of the rms jerk showed a bias. At the early phase, repeat assessment produced a significantly greater value of rms jerk than the baseline assessment.

Table 4: Average values (standard deviation), effect size and p-values of rms jerk and lift duration comparing baseline and repeat assessment of controls

Reliability Comparison										
					Effect size			P-Values		
	Assessment	Early	Middle	Late	Early	Middle	Late	Assessment	Phase	A x P
RMS jerk (cm/sec^3)	Control Baseline (SD)	498.7 (214.8)	825.8 (415.8)	948.5 (464.6)	0.86	0.14	0.26	0.089	0.0001	0.372
	Control Repeat (SD)	681.9 (207.3)	879.7 (259.7)	1062 (378.7)						
Lift duration (sec.)	Control Baseline (SD)	2.02 (0.55)	1.48 (0.41)	1.38 (0.38)	0.89	0.41	0.22	0.0001	0.0001	0.003
	Control Repeat (SD)	1.61 (0.34)	1.34 (0.24)	1.31 (0.29)						

Table 5: Average values (standard deviation) of the performance outcome measures comparing baseline and repeat assessments of controls

Performance Outcome Measures				
Assessment	Weight lifted (kg)	Assessment P-Value	Number of lifts	Assessment P-value
Control Baseline	28.4	0.543	67.5	0.848
(SD)	(14.7)		(8.9)	
Control Repeat	33		67.2	
(SD)	(15.9)		(13.4)	

5.3 JERK

A MANOVA revealed a significant difference between the groups comparing control baseline with patient pre-treatment assessment. Patients performed lifts with a lower amount of rms jerk than controls ($p = 0.002$). A phase effect showed both groups increased rms jerk as the task progressed as shown in Fig. 18. The group-by-phase interaction was significant ($p = 0.002$). CLBP patients increased rms jerk from the early phase to the middle phase of the task and then stabilized from the middle to late phase of the task. Controls increased rms jerk over all phases of the task.

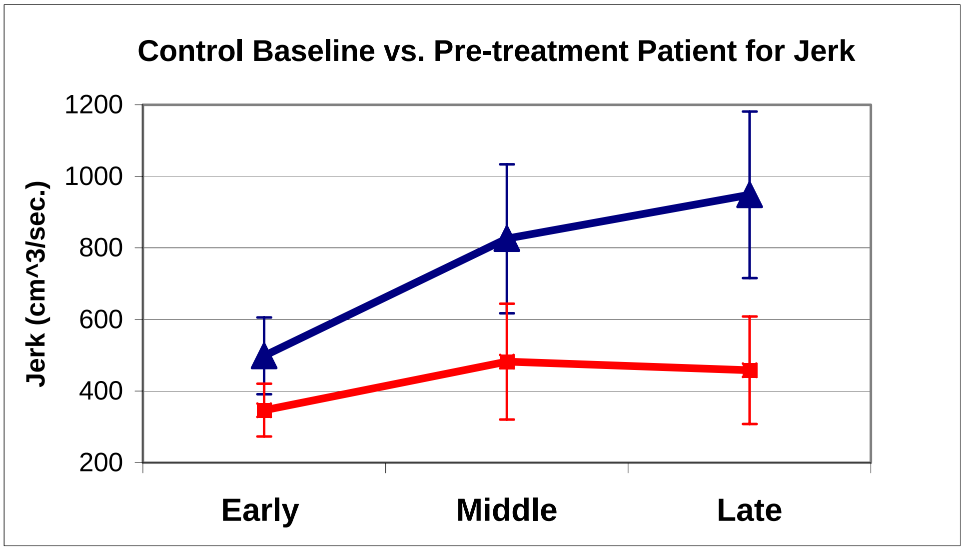

Figure 18: Plot of rms jerk values at each phase comparing controls at baseline assessment with patients at pre-treatment assessment. ▲: Controls at baseline and ■: Patients at pre-treatment

An example of the difference in the magnitude of rms jerk between the groups is shown in Figure 19, which compares jerk for a control to jerk for a patient. The peak jerk magnitude of the control was approximately 1700 cm^3/sec, while the peak jerk magnitude of the CLBP patient was approximately 450 cm^3/sec. The duration of the lift was shorter for the control than the CLBP patient. Lift duration for the control subject was 1.5333 seconds while the patient's lift duration is 2.0667 seconds.

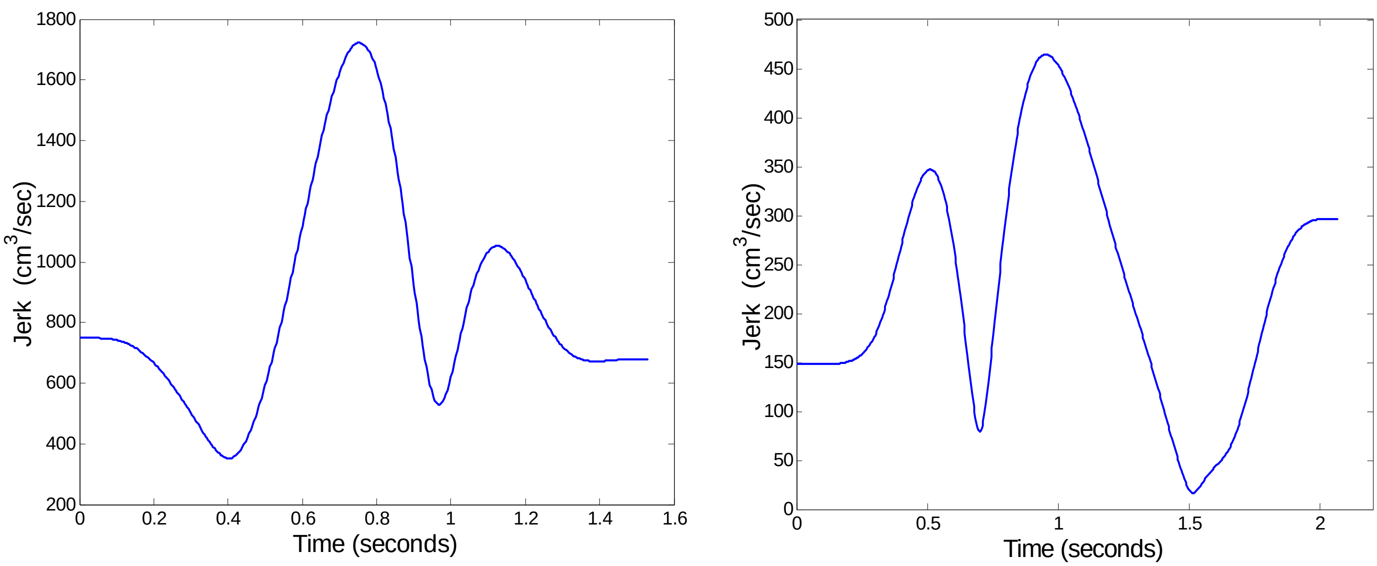

Figure 19: Jerk plotted versus lift time for a control (left) and a patient (right)

A significant difference in rms jerk between patients pre-treatment and post-treatment was found ($p < 0.0001$). In post-treatment assessment, patients performed lifts with greater jerk than in pre-treatment assessment. An effect for phase was significant ($p = 0.011$). Patients increased rms jerk as the task progressed in both assessments. An assessment-by-phase interaction was not significant between the assessments ($p = 0.31$). Post-treatment patients have a greater magnitude of rms jerk but still demonstrate the same changes of rms jerk over task phase as in pre-treatment assessment.

RMS jerk was compared between post-treatment patients and control baseline assessment to determine whether any difference between groups existed after rehabilitation. No significant difference was found for rms jerk ($p = 0.116$). Both phase ($p < 0.0001$) and a group-by-phase interaction ($p = 0.002$) were significant as shown in Fig. 20. Controls increased rms jerk as the task progressed and patients increased rms jerk from early to middle phase and then stabilized from the middle to the late phase.

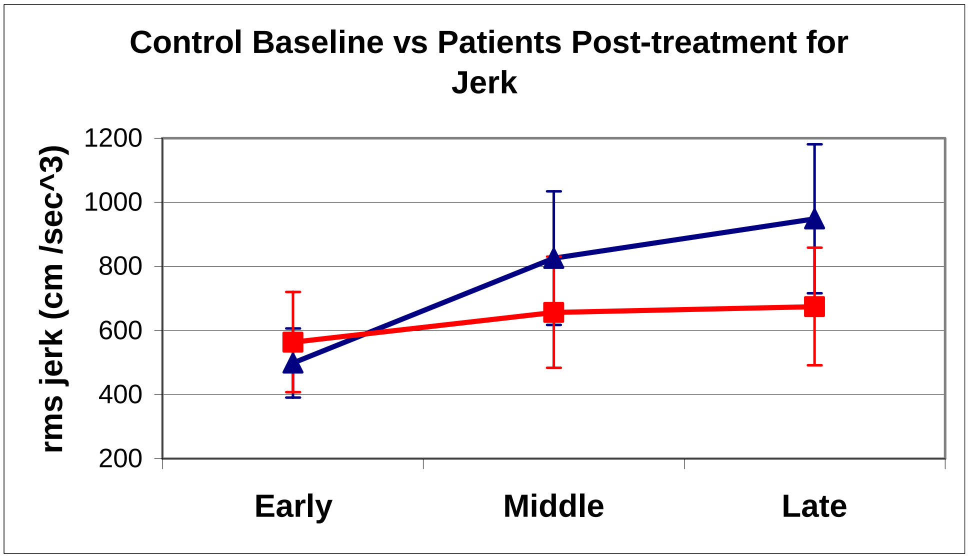

Figure 20: Plot of jerk values at each phase comparing controls at baseline assessment with patients at post-treatment assessment. ▲: Control at baseline and ■: Patients at post-treatment

5.4 PERFORMANCE MEASURES

Weight lifted and lifts completed are the performance outcome measures where only a group difference can be tested for statistical significance. An ANOVA revealed significant group differences for weight lifted ($p < 0.0001$) and lifts completed ($p < 0.0001$). Control subjects lifted more weight and completed more lifts than patients. A significant difference was found for weight lifted between pre-treatment and post-treatment assessment ($p = 0.012$) with patients lifting more weight at post-treatment assessment than at pre-treatment. Lifts completed did not differ between the assessments. Tables 3 and 6 list the mean and standard deviation of the performance outcome measures.

Table 6: Average values (standard deviation) of performance outcome measures comparing assessments of patients

Performance Outcome Measures comparing Assessments				
	Weight lifted (kg)	**Assessment P-Value**	**Number of lifts**	**Assessment P-value**
CLBP Pre-TX	16.9	0.012	41.1	0.183
(SD)	(12.6)		(16.4)	
CLBP Post-TX	21.7		46	
(SD)	(14.2)		(19)	

The performance style measures were tested for statistical differences of group, phase and group-by-phase. A significant group difference comparing patients pre-treatment with control baseline assessment was found for lift duration ($p = 0.015$). Control subjects lifted faster than CLBP patients. Both groups lifted faster as the task progressed demonstrated by a significant phase effect ($p < 0.0001$). The group-by-phase interaction was significant ($p < 0.0001$) and is shown in Fig. 21. Controls decreased lift duration as the task progressed while patients decreased lift duration from early to middle phase but stabilized from middle to late phase. Average values along with standard deviation, effect size and p-values of rms jerk and the performance style measures comparing controls baseline with pre-treatment patients are listed in Table 7.

A significant difference between patient treatment assessments was found for lift duration ($p = 0.018$). In the post-treatment assessment, patients lifted faster than in pre-treatment assessment. Over the task phase, patients at post-treatment performed lifts faster ($p = 0.005$) but no assessment-by-phase interaction existed. Average values along with standard deviation, effect size and p-values of rms jerk and lift duration comparing treatment assessments are listed in Table 8.

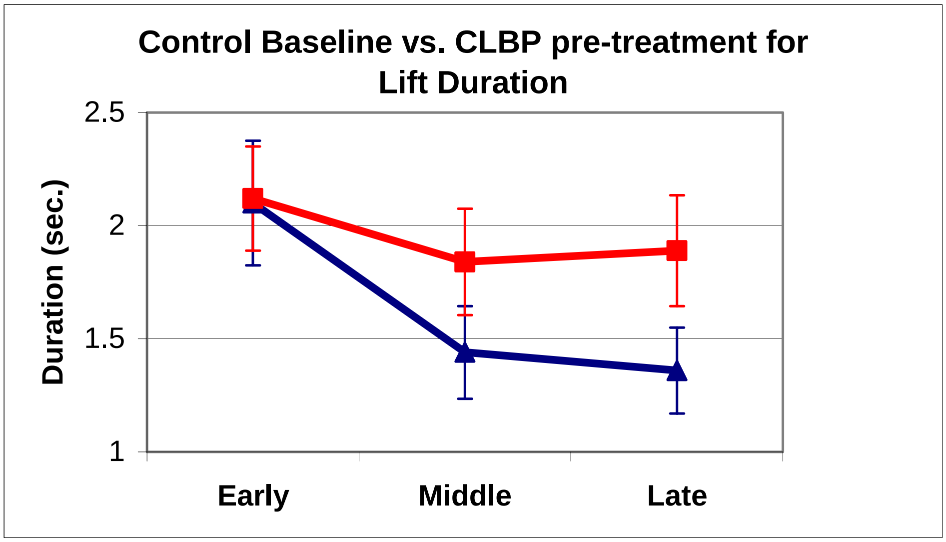

Figure 21: Plot of lift duration at each phase comparing controls at baseline assessment with patients at pre-treatment assessment. ▲: Patients at pre-treatment and ■: Controls at baseline

No significant group difference was found when comparing lift duration between controls baseline and patients post-treatment ($p = 0.767$). A phase effect ($p < 0.0001$) and group-by-phase interaction ($p < 0.0001$) were both found to be significant. The phase effect showed both groups performed lifts faster as the task progressed. The group-by-phase interaction showed control decreased lift duration as the task progressed while patients decreased lift duration from early to middle phase but stabilized from middle to late phase. Average values along with standard deviation, effect size and p-values of rms jerk and lift duration comparing controls baseline with post-treatment patients is listed are Table 9.

Significant group difference was found for starting posture between baseline controls and pre-treatment patients ($p = 0.014$). Controls performed lifts using a more torso lifting style than patients. As the task progressed, starting posture of both CLBP patients and controls became increasingly more torso ($p = 0.039$). No group-by-phase interaction was found for starting posture. Treatment did not have any effect on starting posture. The p-values for assessment, phase and assessment-by-phase interaction were all insignificant.

Midpoint difference compared between controls baseline and patients pre-treatment did not produce any significant findings for group, phase and group-by-phase interaction. A significant difference between groups was found for risetime difference ($p = 0.023$). No phase effect or group-by-phase interaction existed for the risetime difference. Treatment comparison showed no statistically significant difference of assessment, phase and assessment-by-phase interaction of either the midpoint or risetime differences.

Comparison of Figs 18 and 21 suggests that lift duration and rms jerk may be correlated. Controls increased rms jerk and decreased lift duration as the task progressed. Patients decreased lift duration and increased rms jerk from the early to the middle phases and then stabilized from the middle to the late phase for both parameters. Since jerk is calculated as the third derivative of position as a function of time, the interaction observed in rms jerk may be biased by the differences in lift duration. A Pearson correlation confirmed a strong negative relationship between lift duration and rms jerk ($\rho = -0.708$) as seen in Fig. 22. Separating the data by groups showed both groups produced a significant negative correlation between jerk and lift duration (controls $\rho = -0.7$ and patients $\rho = -0.55$). To eliminate the impact of the correlation between rms jerk and lift duration, an analysis of covariance (ANCOVA) was performed, using the early, middle and late lift duration measure for each subject as a covariate and the rms jerk as the dependent measure. The ANCOVA showed that rms jerk still had a significant difference for group, task phase and group-by-phase, when controlling for lift duration as a covariate. The p-values in Table 6 were the values calculated from the ANCOVA.

No significant correlation was found between reported pain severity and jerk of CLBP patients (ρ = 0.01).

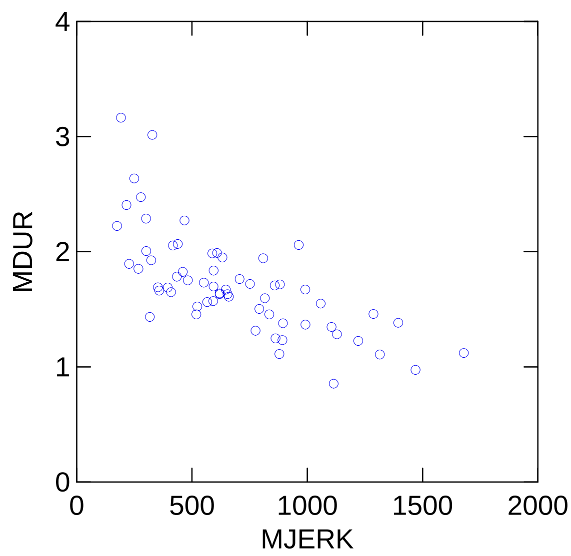

Figure 22: Pearson correlation of mean rms jerk and mean lift duration

5.5 EFFECT SIZE INDEX

Effect size index, which is a measure of standard deviation difference between groups, was calculated for all of the parameters in this study to compare groups or assessments. Effect size was calculated to provide addition statistical information by quantifying how large a difference exists between the groups.

The effect size of the rms jerk comparing control baseline assessment to pre-treatment patients showed a large effect of rms jerk for all phases of the task. Lift duration produced a large effect for the middle and late phases. The effect sizes of starting posture and risetime difference both varied from large to medium effect size depending on phase. The midpoint difference produced a small effect for all phases.

Large effect size for all phases was found for rms jerk and lift duration when comparing pre-treatment with post-treatment assessment. The effect size for starting posture varied from small to medium, and risetime difference varied from large to medium, depending on phase. Midpoint difference produced a small effect size for all phases of the task.

Jerk in baseline controls compared to post-treatment patients revealed a small effect for the early phase, medium effect size for the middle phase and a large effect size for the late phase. Lift duration produced a small effect size for the early and middle phases and a large effect size for the late phase.

Comparison of baseline and repeat assessment of control subjects resulted in a large effect size for rms jerk for the early phase and a small effect size for middle and late phases of the task. Lift duration effect size varied from large at early phase to small at the late phase of the task. Starting posture, midpoint and risetime differences all resulted in small effect sizes for all phases. Effect size indices are located in Tables 3-9. In general, effect sizes for jerk were as large or larger than for the other parameters.

Table 7: Average values (standard deviation), effect size and p-values of rms jerk and performance style measures comparing controls at baseline assessment with patients at pre-treatment assessment.

	Control Baseline			Patient Pre-TX			Effect Size			P-Values		
	Early	Middle	Late	Early	Middle	Late	Early	Middle	Late	Group	Phase	G x P
RMS Jerk (cm /sec^3) (SD)	498.7 (214.8)	825.8 (415.8)	948.5 (464.6)	347.1 (147)	482.6 (324.1)	458.3 (301)	0.77	0.88	1.16	0.002	0.0001	0.007
Lift Duration (sec.) (SD)	2.1 (0.55)	1.44 (0.41)	1.36 (0.38)	2.12 (0.46)	1.84 (0.47)	1.89 (0.49)	0.19	0.83	1.14	0.015	0.0001	0.0001
Starting Posture (SD)	0.227 (0.24)	0.265 (0.26)	0.273 (0.27)	0.061 (0.2)	0.103 (0.27)	0.157 (0.33)	0.73	0.62	0.4	0.014	0.039	0.185
Midpoint Difference (SD)	0.018 (0.047)	0.019 (0.066)	0.026 (0.079)	0.008 (0.041)	0.022 (0.049)	0.016 (0.071)	0.22	0.05	0.13	0.684	0.449	0.146
Risetime Difference (SD)	-0.010 (0.036)	-0.014 (0.038)	-0.014 (0.026)	0.007 (0.026)	0.005 (0.026)	0.011 (0.030)	0.58	0.47	0.92	0.003	0.682	0.576

Table 8: Average values (standard deviation), effect size and p-values of rms jerk and lift duration comparing pre-and post-treatment assessment

Treatment Assessment Comparison										
					Effect size			P-Values		
	Assessment	**Early**	**Middle**	**Late**	**Early**	**Middle**	**Late**	**Assessment**	**Phase**	**A x P**
RMS Jerk (cm /sec^3)	CLBP Pre-TX (SD)	347.1 (147)	482.6 (324.1)	458.3 (301)	2.08	0.76	1.02	0.0001	0.011	0.31
	CLBP Post-TX (SD)	563.9 (313.1)	656.9 (346.9)	675 (367.5)						
Lift duration (sec.)	CLBP Pre-TX (SD)	2.12 (0.46)	1.83 (0.47)	1.85 (0.49)	0.84	0.89	0.63	0.018	0.005	0.473
	CLBP Post-TX (SD)	1.87 (0.52)	1.56 (0.51)	1.65 (0.49)						

Table 9: Average values (standard deviation), effect size and p-values of rms jerk and lift duration comparing controls at baseline assessment with patients at post-treatment assessment.

	Control Baseline			Patient Post-TX			Effect Size			P-Values		
	Early	**Middle**	**Late**	**Early**	**Middle**	**Late**	**Early**	**Middle**	**Late**	**Group**	**Phase**	**G x P**
RMS Jerk (cm /sec^3)	498.7	825.8	948.5	563.9	656.9	675	0.26	0.42	0.62	0.116	0.0001	0.002
(SD)	(214.8)	(415.8)	(464.6)	(313.1)	(346.9)	(367.5)						
Lift Duration (sec.)	2.1	1.44	1.36	1.87	1.56	1.65	0.28	0.18	0.65	0.767	0.0001	0.0001
(SD)	(0.55)	(0.41)	(0.38)	(0.52)	(0.51)	(0.49)						

6.0 DISCUSSION

Jerk in a repetitive lifting task was calculated to describe lifting pattern differences between CBLP patients and pain-free controls. Jerk is the third derivative of position and is a very noisy measure. Substantial smoothing is necessary to obtain useful estimates. A simulation study was conducted to determine which smoothing technique would most effectively smooth subject data. In the simulation study, a synthetic position trajectory to approximate the position trajectories typically observed for subject data was produced from a modified hyperbolic tangent function. The third derivative of the synthetic trajectory provided a known third derivative noise-free approximation. Noise was added to the known synthetic trajectory producing position trajectory signals with SNRs ranging from 40-90 dB. Smoothing methods based on residual analysis and hepatic splines were applied to each noisy trajectory and then the third derivative was calculated. The third derivative from each of the smoothing methods was compared to the noise-free approximation to calculate the MSE and the SNR of the 3rd derivative. The hepatic spline was found to minimize the MSE and maximize the SNR of the 3rd derivative better than the residual analysis. The hepatic spline was therefore used to smooth subject data.

The hepatic spline was fit to subject data, and jerk was calculated for each lift completed. This jerk calculation demonstrated slightly more variability than the synthetic trajectory had predicted even though the data was smoothed. Possible explanations for this variability may be the squaring of the x and y 3rd derivative in the jerk calculation or a result of the subject's lifting technique. Since the cause of the variability is unknown, parameters such as minimum or maximum values of jerk are probably unreliable. The rms amplitude of jerk was used to provide a more stable measure to describe lifting patterns.

RMS jerk, along with several performance parameters, was calculated, and repeated measures MANOVA was used to determine statistical differences between groups and over task phases. Significant differences between the groups were found for rms jerk, lift duration, weight lifted, number of lifts completed, starting posture and risetime difference. Controls performed lifts faster with larger magnitude of shoulder rms jerk than patients. Patients performed fewer lifts than controls and lifted less weight. Controls demonstrated a more torso style starting posture, and patients performed lifts with a more freestyle starting posture. Both controls and patients demonstrated motion in which the midpoint difference is positive, indicating that the hip reached its midpoint before the knees. Controls performed lifts with risetime differences opposite in sign than the midpoint difference. According to Boston et al., this result indicates that both hip and knee angles reached full extension at the same time resulting in a coordinated ending lift [8]. Patients performed lifts with a positive risetime difference and a positive midpoint difference; indicating that the knees straighten before the hip and producing what Boston et al. classified as an uncoordinated ending lift [8].

Significant changes over task phases (early, middle, late) were observed for the parameters of rms jerk, lift duration and starting posture. As the task progressed, both groups increased rms jerk, decreased lift duration and demonstrated an increasingly torso style starting posture. The increase of lifting speed and change to a torso starting posture as the task progressed are consistent with the results of Dolan [16] and Sparto et al. [6] that were reviewed in Chapter 2. A significant group-by-phase interaction was found for rms jerk and lift duration. CLBP patients increased speed and rms jerk from early to middle phase of the task and then stabilized both parameters from middle to late phase. Controls increased rms jerk and lift duration for all phases of the task.

Each of the patients completed a rehabilitation program after the first lifting task assessment. After the rehabilitation program was finished, patients performed the lifting task again to compare lifting patterns pre-treatment with post-treatment assessment. Treatment showed differences in rms jerk, weight lifted and lift duration. Post-treatment CLBP patients performed lifts faster with greater rms jerk while lifting more weight. Significant differences over task phase were found for post-treatment CLBP patients. Jerk and lift duration increased as the task progressed. Assessment-by-phase interaction was not significant for lift duration and rms jerk, indicating that patients post-treatment demonstrated the same changes in these parameters over the task phases as pre-treatment. Jerk and speed increased from early to middle phase and then these parameters stabilize from middle to late phase of the task.

A subset of control subjects completed the lifting task after 3 ½ weeks to assess reliability of the parameters. An intra-class correlation revealed rms jerk to have a moderate reliability. A MANOVA revealed lift duration as the only parameter to reveal a significant difference between the assessments of controls. Controls performed lifts faster during the repeat assessment compared to the baseline assessment. An effect for phase was significant for rms jerk and speed indicating controls increased both parameters as the task progressed. An assessment-by-phase interaction did exist for lift duration. No assessment-by-phase interaction existed for rms jerk or any of the other performance parameters.

In this study, CLBP patient showed a lifting style that was slow and smooth (low jerk) while controls used a faster, more forceful (greater jerk) lifting style. Discrepancies between the lifting styles may be due to the general deconditioning that exists among the CLBP population.

As discussed in Chapter 2, Kankaanpää et al found CLBP patients showed greater fatigability of the gluteal maximus muscle during a flexion-extension task [18]. Oddsson and De Luca found that patients perform at a lower maximum voluntary contraction of paraspinal muscles than controls and the presence of pain causes a redistribution of the activation behavior between synergistic muscles of the lumbar back [40]. Paraspinal back muscles in healthy controls were found to have a greater ability to globally offset local segmental activation imbalances in an isometric contraction than in patients [40]. Newcomer et al. studied the activation of the trunk muscles during platform perturbations and found that absence of muscle activation in the trunk muscles of lower back pain subjects suggests an abnormality somewhere in the neuromuscular system [41]. The lack of muscle strength and muscle activation could explain the inability of CLBP subjects to produce a high rms jerk lift. The greater fatigability of the hip muscle extensors and limited flexibility may be a reason for significant group-by-phase interaction between controls baseline and patients pre-treatment. Patients performed a lifting style where jerk and speed stabilized from middle to late phases of the task after increasing from early to middle phase. The fatiguing muscles could have prevented patients from increasing jerk and speed from the middle to the late phases of the task, which was characteristic of controls in this project.

Another factor that contributes to differences in lifting patterns may be the fear of pain that CLBP patients often experience. Fear of pain causes patients to avoid social and physical activities and may lead to physical and psychological consequences augmenting disability [42]. Vlaeyen et al. found a significant covariation between left paralumbar muscular activity and the pain report, suggesting that the presence of pain results in tensing of the muscles in patients [42]. A study by Waddell found that CLBP patients limit physical function due to pain, which may be associated with muscle guarding and disruption of motion [43]. A high jerk lift would require large spontaneity of motion, and patients' fear of pain would not permit such a lifting style.

A lifting style involving greater rms jerk biomechanically indicates a lift where a subject demonstrates greater spontaneity of motion. Puniello et al found stronger elderly subjects produce more peak jerk than weaker elderly subjects when lifting a box, suggesting that stronger subjects merged the entire lift into one movement while weaker subjects divide the lift into two distinct motions [29]. The lifting study presented here only focused on the lifting of the load and not lowering, so it is only speculation that high jerk lifts result in lowering and lifting as a relatively continuous movement. A control subject using a high rms jerk lifting style would produce a large force at the handle to lift the load. The motion strategy of a high rms jerk lift suggests that the subject has enough muscle strength and endurance to accelerate the load and uses this acceleration as momentum to lift the load resulting in a high jerk lift. The lower rms jerk lift demonstrated by CLBP patients suggests a lifting pattern where subjects use guarded motion possibly fearing injury. This motion strategy suggests that CLBP patients produce a force at the handle that is required to lift the load and not much greater. The motion strategy of patients could be to avoid pain, resulting in reluctance of patients to produce large changes of forces or acceleration.

A comparison of post-treatment patients with baseline controls revealed no overall group difference of rms jerk or lift duration. However treatment did not normalize CLBP patients to perform as controls when examining changes over time. A group-by-phase interaction was found to be significant for both measures. The group-by-phase interaction revealed that controls increase rms jerk as the task progressed, but CLBP patients increased from early to middle and then stabilized from middle to late. The short period of rehabilitation could explain the significant group-by-phase interaction.

The main hypothesis of this study is that rms jerk could discriminate motion patterns between pain-free controls and CLBP patients and could detect differences due to the effects of treatment. Two different statistical methods, MANOVA and effect size, were performed to assess rms jerk as a parameter to detect differences of motion. The MANOVA results showed differences between groups, over task phase and effect of treatment. Effect size was utilized to determine how large a difference between the group means existed. An effect size range of 0-0.32 is small, 0.33-0.55 is medium effect and 0.56-1.2 is a large effect [39]. Jerk was the only parameter studied to produce a large effect size for all phases when comparing baseline assessment of controls versus patients at the pre-treatment assessment. This result suggests that rms jerk had the greatest discriminatory power to detect differences between control at baseline and CLBP patients pre-treatment.

The effect size index was compared between pre-treatment and post-treatment parameters to assess treatment effects. Jerk and lift duration demonstrated large effect sizes for all phases indicating these parameters were able to detect changes due to treatment in this study. The comparison of effect size between post-treatment patients and control baseline assessment revealed a small effect size for rms jerk for early phase, medium effect size for the middle phase and a large effect size for the late phase. The change in effect size from small to large as the task progressed suggests that post-treatment patients started the task with the same jerk values of controls but were unable to continue to perform like controls by increasing jerk during the middle and late phases of the task.

As mentioned earlier, several performance style measures in addition to rms jerk were utilized in this study to describe lifting patterns. Of all of the performance style measures, lift duration detected the most distinct differences between groups, phase and group-by-phase. One may question why calculate rms jerk when lift duration is a less complicated calculation. A significant negative Pearson correlation confirmed that a relationship existed between lift duration and rms jerk when comparing control baseline assessment and patient pre-treatment assessment. The repeat assessment of controls revealed a discrepancy between the measures when compared with baseline assessment. Assessment and assessment-by-phase interaction were significant for lift duration but not for rms jerk. Both lift duration and rms jerk increased from early to middle phase. However lift duration increased for the middle and late phases while rms jerk did not. This finding suggests that lifting speed and jerk did not capture the same motion changes of this task. Lifting speed may be affected by the subject's familiarity to the task. Subjects were unaware of the weight lifted, and the BTE work simulator is uncommon machinery to most controls, so they may be hesitant during the baseline assessment. At the repeat assessment, controls are more comfortable with the lifting task, performing faster lifts. Another major advantage of measuring jerk is that jerk is suggested as providing a biomechanic parameter that describes smoothness, changes of force and acceleration. Lift duration cannot be as directly interpreted in terms of biomechanics.

Flash and Hogan first hypothesized the jerk motion control model of human arm movement in which the model proposed that jerk is minimized [9]. The model suggested that the central nervous system plans movements to maximize smoothness and that practiced and learned motion will result in smoother motion trajectories. Schneider and Zernicke and Young and Marteniuk compared practiced movements with unpracticed movements to determine whether learning would minimize jerk [24] [11]. Both of these studies controlled movement speed by requiring subjects to repeat the task at the same speed demonstrated during the unpracticed trials. The controlled speed eliminated any confounding effects of changing speed could have on jerk. Schneider and Zernicke found for arm motion, practiced trials produced lower jerk [24]. Young and Marteniuk found jerk to increase with practice during a weighted repetitive kicking task [11].

In this project, jerk was utilized as a parameter to detect lifting pattern differences between controls and CLBP patients during a lifting task. The purpose of the project was to utilize jerk as a parameter and not to assess learning of lifting motion. However, if learning can be assumed to occur after the early phase of the task, then motion at the middle and late phases of the task can be considered practiced movements. Lifting speed was not controlled during the task. To eliminate the effects of lifting speed on jerk, an ANCOVA was performed with speed as the covariate. A phase effect was significant for both control baseline and patient pre-treatment assessments. Lifts performed during the middle and late phases produced higher jerk values for both groups than lifts performed during the early phase of the task. Therefore, for this lifting task, learned and practiced motion maximized jerk. This result is consistent with Young and Marteniuk for a repetitive kicking task.

The increase of jerk as the task progressed for the repetitive lifting task violates the minimum jerk model. The minimum jerk model has been supported by studies of planar arm movement studies [24]. However other tasks such as those studied by Puniello and Young and Marteniuk[29] [11], involving motion of multiple body segments, have not followed the model. These results suggest that whether jerk is minimized is dependent on the task. In fact, subjects may attempt to maximize jerk in some tasks. The lifting task used in this project required subjects to lift a relatively large external load, which was not required in studies of Hreljac and Schiender and Zernicke [25] [28] [24]. An external load was present in the studies of Puniello and Young and Marteniuk [29] [11]. To lift an external load, subjects must produce enough force to overcome the weight of the load. Subjects that maximize jerk would generate greater force onto the external load, which possibly make lifting the load easier. Therefore, the minimum jerk model for describing motion may not be valid for tasks requiring the lifting of an external load.

7.0 CONCLUSION

Jerk showed more robust changes between CLBP patients and controls and between patients pre-treatment and post-treatment than any other parameter. Control subjects lifted with greater jerk values throughout the task than patients. Patients increased jerk values after treatment suggesting that treatment improved performance. Even though performance improved, differences between controls and patients still existed suggesting that pain avoidance and muscle deconditioning could limit patients from performing high jerk lifts throughout the task. Jerk was correlated with lift duration but also captured other aspects of lifting that may reflect intersegmental motion. Jerk increased for both groups as the task progressed, which suggests the minimum jerk model may not be appropriate for this task. However, jerk was successful in measuring differences of lifting patterns.

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