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A three-dimensional, dynamic model of the human body for lifting motions

by

Jason Clay Gillette

A dissertation submitted to the graduate faculty
in partial fulfillment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

Co-majors: Engineering Mechanics, Biomedical Engineering
Major Professors: Alison B. Flatau and Timothy R. Derrick

Iowa State University

Ames, Iowa

1999
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This is to certify that the Doctoral dissertation of

Jason Clay Gillette

has met the dissertation requirements of Iowa State University

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For the Graduate College
DEDICATION

This dissertation is dedicated in the memory of Dr. Jeffrey C. Huston, who passed away in the Fall of 1997 after battling esophageal cancer. He was only forty-six at the time and left behind a wife, son, daughter, and a large extended family of former students whose lives he had touched. I met Dr. Huston in the Fall of 1989, when he took the time to talk with me as I found myself beginning to struggle and simply going through the motions in my studies. His thoughtful suggestions led to a critical change in course for me and allowed me to discover an area that I was truly interested in, which was the mechanics of human motion. Dr. Huston cared deeply about the difficulties that the elderly encountered due to falling and gave me the opportunity to be an undergraduate assistant on a project to develop protective hip pads. Though my duties were very simplistic to start with, I began to develop an appreciation for the steps taken during the research process, and my thoughts turned toward graduate school. Dr. Huston was also concerned about the state of engineering education and provided me a further opportunity as a research assistant in the Courseware Development Studio. It was here that I was introduced to philosophical aspects of learning, an experience that has shaped my approach to teaching to this day. Dr. Huston guided me through my Master's research in heart disease, and even though it was not his area of expertise, took a keen interest in my investigation. Dr. Huston supported my decision to go on for a Ph.D. and encouraged me to research a long-time interest of mine, the problem of injuries incurred while lifting. It was Dr. Huston who recommended me for an open slot to teach dynamics, another critical event that has led to my current pursuit of an academic career. Without Dr. Huston's tutelage, my college career would likely have taken a dramatically different direction. It is disappointing to me that I was not far enough along in my research to have Dr. Huston witness one more of his students conclude his studies. Dr. Huston had a special talent of knowing when to push me when I needed to be challenged, and when to praise me when I needed my confidence to be boosted. He is someone that I deeply miss as a teacher, as a mentor, and as a friend.
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ABSTRACT

Lower back pain is prevalent in society and manual lifting has been linked as one potential cause of these types of injuries. Therefore, the 3dLift biomechanical model was developed in this research with the goal of quantitatively analyzing lifting motions. The model divided the body into fifteen segments that were connected by fourteen anatomical joints. During experimental trials, a volunteer subject lifted an object using four different lifting combinations: symmetric leglifts, asymmetric leglifts, symmetric backlifts, and asymmetric backlifts. In order to individualize the 3dLift model, anthropometric parameters were estimated using measurements taken on the subject. During the lifting trials, the subject wore reflective markers placed on anatomical landmarks, the motions of which were tracked by five video cameras. The subject also stood with each foot on a separate force platform that was used to determine ground reaction forces and centers of pressure. Signal processing methods were utilized to predict the marker positions that were obscured during the lifting trials, and digital filtering was implemented to attenuate noise in the data. After reducing the experimental errors, the segment coordinate axes, Cardan angles, joint center positions, and mass center positions were calculated. The changes in the segment orientations with respect to time were then analyzed to determine the three-dimensional kinematics of the segments. Anthropometric, video, and force platform information were combined in equations of motion that were derived to predict the forces and moments occurring at the joints during the lifting motions. A lower body formulation was developed that started with the measured ground reactions at the feet and proceeded through the segments to the T10/T11 intervertebral joint. Similarly, an upper body formulation was derived that began with a known lifted load at the hands and continued through the segments to the same T10/T11 intervertebral joint. While predicting joint forces and moments, the two formulations also served as a means of validating the 3dLift model by comparing the results at the T10/T11 joint. While there is much work yet to be done in this research area, the 3dLift model takes the first steps by developing a systematic methodology for studying lifting motions.
CHAPTER ONE: INTRODUCTION

A wide variety of occupations require manual materials handling, and consequently many workers may be at risk of suffering job-related injuries performing these activities. Data published by the National Safety Council indicate that more work injuries occurred to the back than any other body part from 1987 to 1994 (Mital et al., 1997). Not only are back injuries prevalent, but they can also be disabling to the worker, who may no longer be able to perform the physical requirements of his/her job. As reported by the National Center for Health Statistics in 1985, back impairment was the most common cause of chronic activity limitation in people under age 45 and ranked third in those aged 45-64 (Khalil et al., 1993). Compounding the problems of frequency and severity is the great economic burden to the individual and the economy associated with back injuries. The National Institute of Occupational Safety and Health estimated that the total cost of back injuries for the United States in 1991 was between 50 and 100 billion dollars (Mital et al., 1997). Furthermore, lower back pain is considered to be the most expensive health care problem in the 30-50 age group (Khalil et al., 1993). Statistics such as those listed above reveal that the lower back injury problem is one that has enormous impact on society.

The magnitude of the lower back injury problem has led to a great amount of research in this area. However, when examining the National Safety Council data mentioned above, these efforts do not appear to have significantly reduced the overall incidence of back injuries. Therefore, further research is needed to gain a more complete understanding into how these injuries can be prevented. The amount of applied force required to complete an activity and the posture of the worker during this motion are critical risk factors. Risk assessment is further complicated by the fact that an activity that seems safe to the worker may actually be contributing to cumulative damage to the lower back.

It was the goal of this research to develop a model that accurately predicts the forces and moments at the anatomical joints, with additional focus given to the back. Given a manual task, this model can be used to analyze the required lifting motion and give an indication of whether or not a worker is at unnecessarily high risk of injury. The model can then be used
to test alternative designs to determine if they are less stressful on the joints and therefore less prone to cause injury.

1.1 Sources of Lower Back Pain

The lower back, or lumbar spine, is a complicated anatomical structure consisting of vertebrae, intervertebral disks, bony processes, musculature, ligaments, and the spinal cord (Figure 1.1). There are five lumbar vertebrae, which consist of a hard outer shell of cortical bone and an inner matrix of spongy cancellous bone. Lumbar vertebrae are the most massive of the vertebrae and serve to support the weight of the upper body in addition to protecting the nerves of the spinal cord (Khalil et al., 1993). Intervertebral disks lie between the vertebrae and are composed of the nucleus pulposus, the annulus fibrosus, and the end plates. The disks act as shock absorbers in the back, while allowing motion to occur between adjacent vertebrae. As loading on the lumbar spine increases, the pressure in the inner, fluid-like nucleus pulposus also increases. The outer annulus fibrosus contains concentric bands of collagen fibers and is pushed outward as the pressure in the nucleus pulposus increases. End plates are made up of hyaline cartilage and connect the intervertebral disk to the adjacent vertebrae (Andersson, 1993).

The articular bony processes of adjacent vertebrae mesh with one another to form apophyseal joints that support and stabilize the lumbar spine. Spinous and transverse bony processes protect the spinal cord and provide attachment points for muscles and ligaments (Breakstone, 1992). The musculature of the lower back generates motion of the torso while simultaneously limiting torso movement and stabilizing the lumbar spine. Some of the major muscle groups of the lower back include the abdominal muscles, the iliopsoas muscles, and the erector spinae. The lower back ligaments connect adjacent vertebrae together and help to stabilize the lumbar spine while maintaining flexibility. Although passive in nature, the posterior ligamentous system is considered by some to also play a critical role in lifting (Gracovetsky, 1990). In the lumbar spine, the spinal cord separates into the cauda equina with nerve roots exiting the spinal column between adjacent vertebrae. These nerve roots control the leg muscles, receive sensory information from the legs and hips, and aid in the function of the abdominal organs.
Figure 1.1: Anatomy of the Lumbar Spine and Intervertebral Disks

The most common cause of lower back pain is muscle or ligament strain, which is often caused by excessive lifting, off-balance motion, or overuse/fatigue. Such injuries may result in a reduction in range of motion due to stiffness, weakness, and painful muscular spasms. Another common source of lower back pain is the degeneration of the intervertebral disks, which is most likely caused by a combination of repeated overexertion and physiological changes due to aging. As a result, tears develop in the annulus fibrosus, the nucleus pulposus propagates into the annulus, and the intervertebral disk begins to bulge. A herniated disk that presses on a nerve root will cause back pain and may result in sciatica, which is pain in the back of the leg and foot associated with the compressed nerve (Breakstone, 1992). The nucleus pulposus may also propagate toward the end plates, causing weakening of the junction between intervertebral disk and the vertebra (Andersson, 1993). End plate failure may result in disk herniation into the vertebra or granular ingrowth into the nucleus pulposus. Injuries to the fourth lumbar/fifth lumbar and the fifth lumbar/first sacral intervertebral disks account for 95% of all disk injuries in the lumbar spine (Gracovetsky, 1990).
Intervertebral disk degeneration can lead to further injury in the lumbar spine, such as damage to the articular bony processes. Degeneration at these apophyseal joints may cause a condition called spondylolisthesis, in which vertebrae shift forward and out of alignment with the rest of the lumbar spine. Weight lifters, football players, gymnasts, and javelin throwers are particularly susceptible to this type of injury (Breakstone, 1992). Disk degeneration and/or apophyseal joint deformation may also result in constriction of the nerves known as spinal stenosis. Researchers have attempted with some success to simulate these various types of injuries in a laboratory setting (Adams and Dolan, 1991). Experimental compressive loading has been linked to failure at the end plates and damage to the central intervertebral disks. Combinations of compression and flexion or compression and lateral bending have resulted in disk herniation, although failure due to repetitive loading is difficult to reproduce (Andersson, 1993). It has been further suggested that combinations of compression and torsion cause degeneration of the end plates, the outer intervertebral disks, and the apophyseal joints (Gracovetsky, 1990).

After the onset of lower back pain, 70% of all people can expect to get better within three weeks without specific treatment, and 90% will be free of discomfort in eight weeks (Breakstone, 1992). Unfortunately, lower back injuries have a high rate of recurrence and approximately 60% of those who have recovered will suffer again from pain within the first year after injury. Treatment of lower back injuries ranges from restricting activities, taking medication, and applying heat or cold at home to open-back surgery. Many rehabilitation programs include a regimen of exercises designed to strengthen the back muscles, while avoiding overexertion and potential reinjury of the back. Individuals who do not respond to treatment for their lower back injury may fall into what has been called the cycles of pain (Khalil et al., 1993). The first cycle is referred to as physical deconditioning and involves dealing with the pain by limiting movement and altering normal posture, which may hinder recovery. This may be followed by the second or drug cycle where medications are taken, tolerance to the drugs increases, and higher and higher dosages are required to alleviate the pain. The third cycle is the depression and anxiety cycle, characterized by frustration over the lingering injury and resulting in the person focusing even more on the pain.
1.2 Predicting the Risk of Lower Back Injuries

Upon consideration of the prevalence, recurrence, and depth of suffering caused by lower back pain, there is a clear need for an improved understanding of how to prevent injuries from occurring. There are many approaches to predicting the risk of lower back injuries. These methods can be classified as epidemiological, psychophysical, physiological, and biomechanical.

The epidemiological approach involves using statistics to study the characteristics surrounding cases of lower back pain to determine patterns of injury. Potential personal risk factors identified include strength, age, gender, motivation, training, endurance, health history, and experience (Mital et al., 1997). Furthermore, some possible workplace risk factors are lifting posture, load characteristics, repetitive handling, workplace geometry, workplace environment, and task duration (see also Table 1.1). Such observations lead to general guidelines for preventing lower back injuries such as increasing strength through exercising and avoiding lifting when tense or overtired (Ishmael and Shorbe, 1985). However, guidelines such as lifting with the legs may not be appropriate for every task and may actually increase loading on the back or transfer excessive loads to other joints such as the knees. Epidemiological studies are valuable in pinpointing occupations where lower back injuries are a problem and also in evaluating whether attempts to reduce injuries are successful.

The psychophysical method relies on worker perceptions of the difficulty in completing a specific task to determine the relative risk of injury. For example, psychophysical studies undertaken by the Liberty Mutual Insurance Company resulted in tables of maximum acceptable weights for manual handling tasks (Snook and Ciriello, 1991). Their experiments allowed subjects to alter the weight or force required for certain tasks to the maximum level they could handle without straining or fatiguing themselves. In a variation of this approach called direct estimation, subjects assigned a numerical value in evaluating the strain required to perform standard tasks (Chen et al., 1992). The accuracy of the psychophysical method depends on the subjects being sensitive to the loads being applied at their joints for different motions. When comparing this approach to biomechanical models, the psychophysical
Table 1.1: Personal and Workplace Risk Factors

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Adapted from Mital et al. (1997) and Khalil et al. (1993)

method may result in excessive recommendations for lifting below the knee level (Nicholson, 1989). However, worker perceptions of task difficulty are unquestionably valuable in a qualitative sense to highlight aspects of materials handling that may pose injury risks. Furthermore, psychophysical studies can indicate unfavorable lifting characteristics for workers such as task frequency, load handles, and unstable balance that may be difficult to model.

Physiological methods include measurements of intra-abdominal pressure, oxygen consumption, and lumbar curvature while lifting. Measurements of intra-abdominal pressure are compared to estimated upper limits that can be produced without injury to the lumbar spine. However, the measuring equipment may limit the lifting motions that can be studied (Nicholson, 1989). This method also suffers from controversy over the role intra-abdominal pressure plays in spinal loading. Different researchers have stated opposing views that intra-abdominal pressure decreases spinal compression, has no effect on spinal compression, or actually increases spinal compression (Mital et al. 1997). Oxygen consumption measurements while lifting are compared to the maximum metabolic energy that can be expended without causing overexertion or fatigue. The use of metabolic cost is not appropriate for infrequent lifting tasks such as those studied in this research, but is an
important consideration for high frequency lifting tasks (Kumar and Mital, 1992). Measurements of lumbar curvature have been related to the flexion/extension moments in the lower back through use of experimental data taken from cadaveric lumbar spines (Adams and Dolan, 1991). This method is specific to flexion/extension of the lumbar spine in the sagittal plane and was not applied to lateral bending, twisting, or combinations of these motions. The physiological methods described above all have valid applications, but are more restrictive than the evaluation of generalized lifting that is the objective of this dissertation.

Biomechanical models use laws of motion and mechanical properties to estimate stresses incurred in the body while lifting. Biomechanical models of the body as dynamic, rigid, linked segments were of most interest to this research, and examples of these models are described in Chapter Eight. Other types of biomechanical analyses include the finite element method, electromyography studies, and simulation based back models. The finite element method involves reconstructing the lumbar spine with a geometric mesh and assigning material properties according to the type of biological structure being defined. When using accurate loading inputs, these models can be valuable in predicting which components of the lumbar spine are most highly stressed and are at risk of being injured (Lavaste et al., 1992; Rao and Dumas, 1991; Shirazi-Adl, 1989). Electromyographic models incorporate measurements of electrical activity of selected muscles and relate these signals to the level of tension in the muscle (McGill, 1992). Such studies have the advantage of determining muscle recruitment, but suffer from complex signal to force relationships and difficulty in accessing the deep muscles of the back. Simulation based models of the lower back can provide insight into subjects such as stability of the spinal architecture (Crisco and Panjabi, 1990) and compression of the fifth lumbar/first sacral intervertebral disk (Chaffin and Andersson, 1991).

There have been attempts to combine aspects of the approaches listed above into a single comprehensive model to recommend limits of weights to be lifted by workers. A prominent example of such a model is the Work Practices Guide for Manual Lifting developed by the National Institute for Occupational Safety and Health (Waters et al., 1993). The NIOSH guideline defines safety limits based on object horizontal and vertical location, object vertical travel distance, angle of twist, frequency of lifting, and type of coupling. While such a model
is commendable, it is intended to apply only for constant velocity lifts and simply looks at the initial and final positions of the lifting object. The model developed in this research (subsequently called 3dLift) differs from the NIOSH guidelines in that it accounted for the effects of segment accelerations, which may be an important factor during lifting. In addition, the 3dLift model examines intermediate positions during the lift, and can therefore differentiate between complex lifting motions. Restrictions of the 3dLift model included low frequency lifting, no slipping between the hands and the lifted load, and ground contact by the feet within the force platform surfaces (see Chapter Three). The 3dLift model was intended to give an accurate prediction of loading in the back, which can then be further analyzed into individual muscle forces or disk compressions if desired.

1.3 An Example: Railroad Yard Workers

Workers performing manual lifting can be witnessed in a wide variety of occupations such as department store stockers, airport luggage handlers, and waiters and waitresses. Railroad yard workers are an example of an occupation where biomechanical modeling could make valuable contributions. Statistics from 1979-1986 indicate that yard workers had the highest lost day injury rate and severity rate in the railroad industry, while accounting for 81 million dollars in claims and suits (Kuciemba et al., 1988). Yard workers perform strenuous duties including mounting and dismounting cars, coupling and uncoupling cars, throwing hand switches, and setting and releasing hand brakes. Of these tasks, throwing hand switches is of particular interest to this research since it is a lifting activity and accounts for over 15% of lost day injuries to yard workers (Figure 1.2). Injuries related to throwing hand switches are mostly due to overexertion, with 43% of these injuries occurring at the lower back (Page et al., 1990). Some of the typical types of injuries reported include back sprains, upper extremity strains, injuries to the knees and feet, and hernias. This is an example of how epidemiological data can be used to identify a task such as hand switch throwing that poses a risk of injury to workers.

Throwing hand switches involves lifting a lever at ground level and rotating it through a 180° arc to change the alignment of track at an intersection. Hand switch operation is a frequently performed duty, with 60 to 70 switches thrown by a yard worker on an average
day (Kuciemba et al., 1988). Hand switch maintenance is an important factor, since throwing well maintained switches is considered an easy task by yard workers and throwing poorly maintained ones a difficult task. Other factors that have been cited as contributing to difficulty of operation are restricted throwing postures, poor weather, switch stand age, and defective switch stand equipment. Therefore, psychophysical evidence also supports the statistical data in linking operation of hand switch stands to the risk of back injury.

Throwing of hand switches has been analyzed by the Association of American Railroads (AAR) using the NIOSH model (McMahan, 1984), but the results should be viewed with a measure of caution. Because this motion requires lateral bending, twisting of the torso, and is dynamic in nature, it violates several of the assumptions outlined in the NIOSH guidelines. The AAR research of hand switch operation was later extended by extrapolating the NIOSH model to include asymmetric postures, but the analysis remained static in nature (Page et al., 1990).

The biomechanical model proposed in this research can be a valuable tool in studying a task like throwing hand switches because of its capability to analyze three-dimensional dynamic motion. By using the 3dLift model, joint forces and moments under specified lifting conditions can be predicted and compared with joint loading after lifting characteristics have been altered. For example, new designs of the hand switch stands could be tested before they are implemented to determine if their operation reduces loading on the joints. In addition, different lifting styles could be studied to determine whether or not a certain method could be recommended to yard workers to potentially reduce injuries. Switch stand maintenance appears to be a critical factor. Furthermore, the author has witnessed the substantial increase in effort required to throw a hand switch that is out of alignment. The 3dLift model could be used to help set limits to forces required for throwing switches as part of a maintenance schedule for adjustment, lubrication, and cleaning of switch stands (Page et al., 1990). Some yard worker activities may not be easily changed, and strength testing has been suggested to ensure that a person is capable of the required duties (McMahan, 1984). Along these lines, prospective workers could perform a series of simulated tasks that were determined by the 3dLift model to have similar requirements to those encountered on the job.
1.4 Dissertation Overview

It was the goal of this research to develop the 3dLift biomechanical model as a means of analyzing three-dimensional lifting motions. In order to study this problem, this dissertation outlines the experimental and theoretical methods incorporated into the 3dLift model. To give a quantitative measure for the amount of loading on the body, the model estimated joint forces and joint moments that were occurring during lifting motions. The model is validated by comparing the T10/T11 joint force and joint moment predictions between a lower body and an upper body formulation. Ultimately, it is hoped that the 3dLift model will prove to be a valuable analysis tool toward the prevention of injuries.

This chapter discussed the prevalence of lower back injuries incurred while performing manual lifting, which indicated the need for further research into the cause of such injuries. It also described a number of methods that researchers have previously used to study lifting motions.

Chapter Two outlines the experimental protocol followed during the lifting trials, including the general lifting motions that were analyzed. The video tracking cameras and force platform equipment used to collect input data are also described.

Next, Chapter Three considers individual anthropometry, including a set of subject measurements and the placement of reflective markers on anatomical landmarks. Based on these measurements, estimations of segment masses and moments of inertia are outlined.

Extrapolation and interpolation methods are considered in Chapter Four as a means of dealing with obscured video marker data. Digital signal processing is described as a means to reduce erroneous noise in the experimental measurements.

Chapter Five focuses on the transformation of video marker data into body segment orientations using triad equations and Cardan angles. In addition, joint center positions and mass center positions are calculated using video marker positions and anthropometric equations.

Three-dimensional kinematic equations are developed in Chapter Six to determine angular velocities, angular accelerations, and linear accelerations of the body segments. Kinematic results for body segments of interest are presented and compared for the variety of lifting motions analyzed.
Chapter Seven details the calculation of ground reaction forces and centers of pressure from force platform data and several simple accuracy tests are described. The integration of force platform data with video-based position data is also outlined.

The experimental and theoretical considerations from Chapters Three through Seven are combined in Chapter Eight as equations of motion. A lower body formulation using force platform data as input and an upper body formulation using known lifted weights as input are detailed. Joint forces and joint moments are reported for the joints of interest and differences between the studied lifting motions are illustrated.

Chapter Nine compares the results of the lower body formulation and the upper body formulation at the T10/T11 intervertebral disk as a means of validating the 3dLift model. Finally, conclusions regarding the 3dLift biomechanical model are discussed in Chapter Ten, and future areas of investigation in this area of research are recommended.
CHAPTER TWO: EXPERIMENTAL SETUP

As a means of experimentally examining lifting motions, a set of lifting trials is introduced in this chapter. The video tracking equipment used to measure the movement of reflective markers placed on the lifting subject is also described. In addition, the force platforms used to measure the subject ground reaction forces during lifting are detailed.

Researchers have used a wide variety of methods to study back injuries, and many have designed experimental lifting trials to gain insight into this problem. Previous investigations have focused on different aspects of lifting motions and have also utilized numerous parameters to evaluate the risk of injury. For example, Patterson et al. (1987) studied the influence of lifting experience, load magnitude, and load knowledge on joint forces and moments at the fourth/fifth lumbar disc. As another example, Bush-Joseph et al. (1988) examined the effects of lifting speed and lifting method on moments at the fifth lumbar/first sacral intervertebral disc (L5/S1). Approaching the problem from a slightly different perspective, Gracovetsky (1990) measured changes in lumbar flexion and lifting speed with increasingly heavy loads. Adams and Dolan (1991) also investigated lumbar flexion along with moments at the L5/S1 disc in relation to an object's weight and bulkiness. Using mechanical work as a means of evaluation, Gagnon and Smyth (1991) studied both lifting and lowering from a variety of heights with different loads. Lindbeck and Arborelius (1991) analyzed lifting techniques and lifting speeds while calculating joint moments, L5/S1 disc compression, and individual segment inertial contributions.

In addition to using predictions of L5/S1 disc compression, Waikar et al. (1991) evaluated a series of lifting tasks based on individual subjective estimates of stress at the lower back. McGill (1992) determined L5/S1 disc moments during lifts involving lateral bending and further partitioned the moments into disc compression, muscle forces, and ligament forces. The effect of lifting primarily with the back (backlift) as opposed to lifting primarily with the legs (leglift) on the joint moments of the L5/S1 disc and the joints of the lower extremity was evaluated by Toussaint et al. (1992). Dolan and Adams (1998) investigated the effects of repetitive lifting and fatigue on lumbar flexion, L5/S1 disc moments, and L5/S1 disc compression. Along with analyzing L5/S1 disc moments, Kingma
et al. (1998) studied pelvic twist as it related to back motion during asymmetric lifting. From the experimental trials listed above, it becomes apparent that although a wide variety of conditions and variables for analysis are used to investigate lifting, there is a common goal of injury prevention. As part of this research project, two different lifting motions and two different lifting styles were studied using joint forces and joint moments as a predictor of the severity of the lift. These experimental lifting trials were chosen to have distinct differences, but to be simple enough to aid in validating the 3dLift biomechanical model.

2.1 Lifting Trial Protocol

A male volunteer with no history of back injury participated in the lifting trials after signing an informed consent to participate in research (see Appendix A). Prior to the start of the study, the subject went through an orientation session to familiarize him with the lifting motions and the equipment used during the data collection. This session provided the subject an opportunity to ask questions and obtain further information about all aspects of the study. Before performing the lifting trials, a set of anthropometric measurements were taken on the subject to individualize the segment parameters of the model as described in Chapter Three. A set of retroreflective markers were then attached to the skin of the subject with double-sided adhesive tape to highlight anatomical landmarks as also listed in Chapter Three. The subject was asked if he was uncomfortable with any of the proposed anthropometric measurements or marker placements before they were performed. During the lifting trials, the positions of the markers were tracked by four real-time cameras using a Peak Video and Analog Motion Measurement system (Peak Performance Technologies Inc., Englewood, CO) as described later in this chapter. The subject stood with each foot on a separate force platform to measure the ground reactions and centers of pressure (see Chapter Seven) while performing the lifting motions.

The subject was asked to perform lifting trials consisting of two different lifting motions (symmetrical and asymmetrical) and two different lifting styles (leglift and backlift). The order of the lifting trials was symmetrical leglift, asymmetrical leglift, symmetrical backlift, and asymmetrical backlift, which were then repeated four times for a total of sixteen trials. Multiple trials of the lifting tasks were executed to ensure that the experimental data showed
repeatability for each lifting motion and each lifting style. A resting period was allowed between lifting trials, if the subject desired, in order to avoid any effects of fatigue that might alter the lifting motions. The object to be lifted was a crate containing free weights. The subject was allowed to warm up and choose a weight that he felt was comfortable and could be safely lifted. This lifting object resembled a milk crate with the side meshing removed to improve visibility of the reflective markers for the video system. The crate was 37.3 cm wide, 33.3 cm deep, 28.2 cm tall, and had handle heights that were located 21.7 cm from the bottom of the crate. The subject was limited to choosing a lifting weight that was equal to or less than fifteen percent of his body weight in order to minimize the risk of an injury occurring during the study. The amount of weight that the subject lifted (12.2 kg) was similar to what might be experienced in a job that required lightweight manual materials handling.

One motion was symmetric in nature and involved lifting the crate from ground level (at handle height) in front of the body to waist level in front of the body. Depending on the lifting style being used (see below), such a motion may require flexion/extension of the back, hips, knees, and/or ankles. The other motion was asymmetric and consisted of lifting the crate from ground level in front of the body to waist level on one side of the body along a diagonal path. In addition to motions required in the symmetric lift, the asymmetric lift also involved twisting and lateral bending at the hips and back. Lateral bending and twisting are motions that were of interest since they have been studied less than flexion/extension and because of their role in potentially increasing the risk of injury during lifting. One lifting style was the leglift, where the subject initiated the lift in a squatting posture and concentrated on lifting primarily with their legs and keeping their back straight (Figure 2.1). The leglift involved little torso motion and more hip, knee, and ankle motion than the backlift. The second lifting style was the backlift, in which the subject began in a stooped position and focused on lifting primarily with their back while keeping their legs approximately straight (Figure 2.2). Therefore, the backlift relied mainly on torso motion with little knee and ankle motion. These two lifting styles were chosen to test the effect of different postures on the loading at the joints, especially in the back.
2.2 Video-Based Kinematic Measurements

Three-dimensional motion measurements are used to analyze motion in many diverse situations such as injury prevention, orthopedics, and rehabilitation. Many investigators research movements of interest using video-based systems to track the positions of a variety of markers placed on the body. For example, lifting motions were studied by Frigo and Pedotti (1993) using an automatic motion analyzer (ELITE) and retroreflective markers, by Toussaint et al. (1992) using a sixteen millimeter camera and reflective markers, by Lindbeck and Arborelius (1991) using an optoelectric camera (Selspot II) and infrared light emitting diodes, and by Patterson et al. (1987) using a sixteen millimeter camera and theatrical greasepaint markings that were manually digitized. Another common approach involves attaching strain gage, piezoresistive, or piezoelectric accelerometers to a body segment and
measuring accelerations (Nigg, 1994). Such an accelerometer system was used by Cook et al. (1990) to measure vertical accelerations of a lifted box. Furthermore, biomechanics research laboratories may develop specialized instrumentation to expedite kinematic measurements. Ferguson et al. (1992) and Marras (1993) tested subjects performing lifts while wearing a portable exoskeleton called the lumbar motion monitor (LMM). Instead of measuring unknown kinematic values, Mirka and Marras (1993) developed a system powered by a Kin/Com dynamometer to control the velocities and accelerations achieved by the subject while lifting.

Three-dimensional kinematic data for this research were obtained using a Peak Video and Analog Motion Measurement System. Many lifting models restrict analysis to the sagittal plane and assume that the movement is symmetrical, considering only two-dimensional flexion and extension of the back. Three-dimensional motion is analyzed throughout this
Asymmetric lifting motions are common, especially tasks that involve twisting motions of the back. Four 120 Hz real-time cameras were used to film the subject, with one camera viewing the subject from the front, one camera viewing the subject from the rear, and the other two cameras approximately thirty degrees to the left and right of the frontal camera. Such a setup is needed because each marker must be seen by at least two cameras to reconstruct three-dimensional coordinates. Allowing markers to become obscured during a trial was avoided if possible, since they must then be manually digitized, which is time consuming, or interpolated, which is likely less accurate. The subject wore retroreflective markers placed on anatomical landmarks, with groups of two or three markers used to define the body segments (see Chapter Three). Each camera lens has infrared lights positioned around it and contains a filter to reduce wavelengths entering the lens from sources other than marker reflections (Peak Performance, 1997).

The real-time cameras have a fixed focal length, so they were positioned where the markers filled as much of the screen as possible without any of them leaving the field of view during the lift. In addition, the camera views were also approximately centered upon the subject's midsection, resulting in the cameras being placed sixteen feet away from the lifting subject. One reconstruction method involves using the direct linear transformation (DLT), which allows two-dimensional coordinates from multiple cameras to be combined into three-dimensional data:

\[
x_{ij} = \frac{A_j x_i + B_j y_i + C_j z_i + D_j}{E_j x_i + F_j y_i + G_j z_i + 1}, \quad y_{ij} = \frac{H_j x_i + J_j y_i + K_j z_i + L_j}{E_j x_i + F_j y_i + G_j z_i + 1},
\]

where \(x_{ij}\) and \(y_{ij}\) are the coordinates of marker \(i\) seen from camera \(j\); \(x_i\), \(y_i\), and \(z_i\) are the transformed coordinates of marker \(i\); and \(A_j\) through \(L_j\) are calibration constants for each camera \(j\) (Allard et al., 1995). For this research, each real-time camera had a linearization file that corrected for lens distortion and perspective errors as a marker moved throughout the field of view (Peak Performance, 1997). Calibration of the cameras involved filming a four-point static reference frame and a dynamic two point wand that was manually moved and rotated throughout the trial volume. Such a process provided a series of coordinate data that, along with the linearization file, determined the video coordinate system and the camera calibration constants. These static and dynamic reference points were distributed to cover the
entire space where movement was to occur to avoid potentially inaccurate extrapolation beyond the calibration volume. It was also critical that the measurement of the calibration frame dimensions and wand length be highly accurate to avoid coordinate reconstruction errors.

Automatic data acquisition of the marker positions was utilized with the real-time cameras to greatly increase the speed of digitization and to increase the accuracy of the coordinate values over manual methods. To facilitate automatic digitization, the video cameras were adjusted to provide maximum contrast between the markers and the background by varying the f-stop. A larger f-stop setting indicated a smaller aperture diameter with respect to the focal length and consequently less light entering the camera lens (Winter, 1990). In addition, the digitizing software allowed the threshold of light it would recognize to be varied to account for different lighting conditions and to increase the contrast of the markers. It was also helpful in terms of marker contrast to have a solid, dark background absent of any reflective or light-emitting objects in the field of view. The cameras were synchronized by sending a video signal from a designated master camera to the Gen-lock inputs of the three remaining slave cameras (Peak Performance, 1997). Force platform signals (Section 2.3) were also sampled at 120 Hz and synchronized with the video camera data using a voltage pulse that was detectable by both systems. In addition, a fifth camera was used for filming video of the lifting subjects, which was useful in giving a visual display to go along with the analytical results.

As interpreted from the user's guide (Peak Performance, 1997), the digitization parameters were set to recommended values of 0.02 for predictor error, 0.0045 for maximum residual, 50 for acceleration factor, and 0.001 for noise factor. In addition, the maximum join gap was set to 5, the maximum valid path to 5, and the extrapolation points to 5. After automatic digitization, the tracked paths were manually assigned to their corresponding markers. Marker identification was aided by looking at a combination of digitization planes and by viewing the video taken by the fifth camera. The Peak Performance software was also set to initially filter the raw three-dimensional marker coordinates at ten Hertz before exporting the data to a text file. This cutoff frequency was chosen for initial filtering since normal human motions were expected to occur at 10 Hz and below (Winter, 1990). These
text files were then input into the 3dLift biomechanical model, which was a FORTRAN program written by the author. The 3dLift program then performed further signal processing as needed on the video data including extrapolation, interpolation, and digital filtering (see Chapter Four).

2.3 Force Platform Measurements

Force platforms were used to measure the ground reaction forces developed between the bottom of the subject's feet and the top of the force platforms during lifting activities. In the 1930's, Elftman (1939) performed pioneering work using a force platform to study ground reactions during human locomotion. Since then, force platforms have been used for a wide range of biomechanics applications, including postural stability (Starck et al., 1993), normal walking (Buczek and Banks, 1996; Cavagna, 1975), and pathological gait (Khodadadeh and Welton, 1992). Platforms have been mounted under the belt of a treadmill to analyze gait (Kram and Powell, 1989) and covered with carpet to study walking (Crowe et al., 1996). Researchers have even sealed and placed force platforms underwater to examine swimming turns (Blanksby et al., 1995). Commercially available force platforms can be purchased from companies such as Kistler Instrumentation Corporation (Amherst, NY), Advanced Mechanical Technology Incorporated (Newton, MA), and Bertec Corporation (Worthington, OH). Custom-built force platforms have also been utilized by researchers (Calder and Smith, 1987), and have been designed to be relatively inexpensive, simply constructed, and transportable.

Most platforms measure force by way of either strain gages or piezoelectric crystals that change output voltage when subjected to deformations. The force platforms used in this study were an AMTI OR6-6-2000 and an AMTI OR6-7-2000 (Advanced Mechanical Technology, Inc., Newton, MA), which utilized strain gages. Strain gages were well-suited for measuring the ground reactions, because of their static and dynamic capabilities, good linearity, and the ability to maintain accuracy within five percent if properly used (Nigg, 1994). In addition, the strain gages were configured within the AMTI force platforms to achieve low levels of signal crosstalk, temperature sensitivity, and voltage drift. Crosstalk would occur when a strain gage detected deformations from forces other than those in the
direction intended for measurement. For example, a strain gage positioned to measure horizontal forces may produce erroneous output in response to deformations that are caused by vertical forces. Temperature compensation was essential since the electrical output to mechanical input ratio of a strain gage is altered with changes in temperature (Measurements Group, 1988). As configured for the lifting trials, the sensitivity limits for the AMTI platforms were approximately 0.4 N (0.08 lb) for horizontal forces and 1.4 N (0.3 lb) for vertical forces.

Proper installation of the force platforms was critical to ensure accurate measurements of ground reactions (AMTI, 1991). The platforms were rigidly bolted onto aluminum mounting plates to prevent the introduction of erroneous moments and forces that would occur if they moved with respect to the floor. The mounting plates were permanently affixed in a concrete pit eighteen inches deep to isolate the force platforms from the building foundation and prevent surrounding vibrations from being detected. The mounting plates were leveled by adjusting set screws and adhered to the concrete pit using epoxy. Leveling of the platforms was an important consideration to properly account for the effects of gravity. In addition, a wooden frame was constructed to surround the force platforms and provide a smooth transition from floor to platform surfaces. The surfaces of the force platforms were allowed clearance from the wooden frame to prevent the development of non-physiological contact forces.

The force platforms and their amplifiers were isolated at an outlet from other equipment during the lifting trials. Platform output voltages were low (microvolts), and thus stable amplification and controlled excitation voltages were critical for accurate data collection (Measurements Group, 1988). Initially, when one of the force platforms was plugged in at the same outlet with a computer, a monitor, and a VCR, unsteady voltage levels were witnessed during calibration trials. One occurrence of these problems is shown in Figure 2.3, which plots voltage readings from the force platform when it was unloaded. The voltage levels were expected to remain constant as it sat idle, but a voltage drop corresponding to 6.5 lb was seen in the vertical direction, with voltage shifts in all six channels. Such errors proved troublesome, because their timing appeared random and at low enough frequencies that signal processing was not an effective solution. However, when the force platform was
separated from these other electrical devices, this voltage problem disappeared. Other potential causes of unsteady voltages included a faulty amplifier or the operation of heavy-duty appliances such as elevators, air conditioners, or dryers in the vicinity.

To ensure that the forces and moments were measured from a consistent baseline, the force platforms were zeroed. Potentiometers located on the amplifier were manually adjusted with a screwdriver until none of the channel indicators were lit. More precise zeroing was achieved by taking readings prior to the lifting trials and subtracting these preloaded values from the loaded values (see Chapter Seven). To get an indication of how much the voltage readings drifted with time, data were collected from the unloaded force platform every five minutes for eight hours after it was turned on (Figure 2.4). Judging from the random nature of the early readings for both forces and moments, it appeared that the force platform required at least one hour to balance itself after it was switched on. After the initial hour of warm-up, the drift became more predictable and remained less than 3.5 Newtons for forces and 0.5 Newton-meters for moments over the eight hour period. Based
Drift in Force Readings

Drift in Moment Readings

Figure 2.4: Force Platform Voltage Drift

on these results, the force platforms were left on continuously within the experimental setting to retain equilibrium. Even though only minor drift was expected, readings were taken with the platform unloaded before and after the lifting trials to minimize potential zeroing errors.

In order to study lifting motions and gain insight into related injuries, a set of
experimental lifting trials was described in this chapter. The lifting trials included two different types of lift, symmetric and asymmetric, and two different lifting styles, leglift and backlift. Real-time video cameras were introduced as a means of tracking reflective markers as the subject performed the trials. Force platforms were discussed as a means of determining the ground reaction forces that are occurring between the floor and the subject's feet during the lifting motions. In the next chapter, the initial steps are taken in the 3dLift biomechanical model by quantifying the physical dimensions of the subject using anthropometry. Anthropometric measurements, marker placement, and estimation of segment masses and moments of inertia are outlined. The processing of the digitized video data and its conversion into kinematic values for the body segments are the focus of Chapters Four through Six. The calculations involved in finding the magnitude of the ground reaction forces along with the center of pressure at each foot are listed in Chapter Seven.
CHAPTER THREE: SUBJECT ANTHROPOMETRY

In order to increase the accuracy of the 3dLift biomechanical model, subject anthropology was used to take into account the different body shapes and sizes of individuals. Measurements taken of the subject, along with markers tracked by video and referenced to anatomical landmarks, allowed estimations of segment parameters to be determined.

Many anthropometric prediction models have been derived from experimental measurements obtained during cadaver research, and several of these studies will be described. Dempster (1955) investigated the center of mass, weight, and moments of inertia for the body segments relative to anatomical landmarks of eight unpreserved cadavers. Clauser et al. (1969) studied similar segment characteristics, but used thirteen preserved cadavers, which allowed a sample selection from a larger population. Chandler et al. (1975) also reported on these segment parameters for six cadavers, and in addition, located the joint centers of rotation by dissection through segmentation planes. Clarys and Marfell-Jones (1994) focused on measuring segment masses of six cadavers, along with the mass components attributed to skin, adipose tissue, muscle, and bone. The number of invasive cadaver studies has likely been limited due to cost, experimental difficulties, and the level of surgical expertise that is critical in such investigations.

Huang and Suarez (1983) calculated masses and moments of inertia of the head, neck, and trunk regions of a young female using computerized tomography. Other researchers have investigated non-invasive techniques, not only to avoid the complexities of invasive cadaver studies, but also to determine segment parameters of live subjects. Pearsall et al. (1996) also used computerized tomography to determine the mass and moment of inertia at each vertebral level of the trunk using four subjects. With computerized tomography (CT), x-ray cross sections through the body are taken, from which tissue boundaries are imaged and densities assigned according to the CT pixel intensity. Zatsiorsky and Seluyanov (1983) determined the center of mass, weight, and radii of gyration for body segments of 100 subjects using a gamma scanner technique. Jensen (1993) estimated body segment masses and moments of inertia of at least sixty-five subjects of varying ages in multiple studies using
an elliptical zone method. Other examples of experimental methods include moment tables, the quick release method, relaxed oscillation method, and magnetic resonance imaging (Nigg, 1994). Many of these techniques are cost prohibitive and/or difficult to perform, and thus their predominant value is in establishing databases of segment parameters that can be used in further studies.

Adjustments, regression equations, and mathematical models have been based on studies such as those mentioned above to extrapolate the segment parameters to random subjects. Hanavan (1964) developed a model to calculate segment masses, center of masses, and moments of inertia of fifteen body segments based on twenty-five subject measurements. Although composed of simple geometric solids with constant density, researchers have claimed the model to generally predict moments of inertia within ten percent. Hatze (1980) presented a model to estimate the inertial parameters of seventeen body segments of nonlinear density derived from 242 subject measurements. Hinrichs (1990) adjusted the Clauser et al. (1969) segment center of mass data to reference joint centers rather than bony landmarks to make it simpler to use. Vaughan et al. (1992) developed regression equations based on the results of Chandler et al. (1975) to predict lower extremity segment masses, mass centers, and moments of inertia. With intentions similar to Hinrichs, de Leva (1996a) revised the Zatsiorsky and Seluyanov (1983) segment inertia parameters to reference joint centers as opposed to bony landmarks. In addition, de Leva (1996b) further analyzed the Chandler et al. (1975) measurements in order to model segment joint centers as simple mechanical equivalents.

3.1 Subject Measurements

The 3dLift biomechanical model divides the human body into fifteen segments as shown in Figure 3.1, with the initial intent to have at least three markers per segment. With these segment definitions in mind, a set of anatomical landmarks was derived to form a basis for subject measurements, marker placement, and segment parameters (Cappello et al., 1995; de Leva, 1996a, b). In order to make comparisons between subjects and to make results repeatable for an individual, the intent was to make the landmarks easy to find on the body. For the feet, the medial edge of the first metatarsal, the lateral edge of the fifth metatarsal,
1. Right Foot
2. Right Calf
3. Right Thigh
4. Left Foot
5. Left Calf
6. Left Thigh
7. Lower Torso
8. Upper Torso
9. Right Upper Arm
10. Right Forearm
11. Right Hand
12. Left Upper Arm
13. Left Forearm
14. Left Hand
15. Head

Figure 3.1: Segments of the 3dLift Biomechanical Model

and the distal fibula at the ankle (lateral malleolus) were located. On the calf and thigh segments, the medial tibia at the knee (medial tibiale), the lateral tibia at the knee (lateral tibiale), and the external prominence of the greater trochanter were identified. As part of the lower torso, the right and left superior iliac spines and the lower boundary of the sternum (substernale) were located. On the upper torso and head, the upper notch of the sternum (suprasternale), the spinous process of the seventh cervical vertebrae (cervicale), and the top of the head (vertex) were found. For the upper extremity, the posterior edge of the scapula at the shoulders (acromion), the lateral edge of the radius at the elbow (radiale), the lateral tip of the radius at the wrist (stylion), and the lateral edge of the second knuckle were identified.
After locating the anatomical landmarks, a series of anthropometric measurements were performed on that subject and used to estimate segment parameters. These measures were adapted from Clauser et al. (1969) and Chandler et al. (1975) and appear in Table 3.1, along with a short description. Subject weight was taken using a standard medical scale in the same clothing worn during the lifting trials, with the male subject wearing biking shorts. Biking shorts were chosen to be snug fitting, since the right and left trochanter markers were placed over the garment. Anatomical landmark heights were measured vertically from the ground using a sliding caliper while the subject stood with arms at his sides and legs slightly apart. Circumferences were determined with a measuring tape positioned perpendicular to the longitudinal axis of the segment. Finally, breadths and depths were taken with a beam caliper oriented so that a line connecting the endpoints would be perpendicular to the segment longitudinal axis. Measurements were taken twice and taken again if they did not agree within two millimeters. The number of measurements was intended to give a comprehensive database of the subject's anthropometry, while requiring a relatively short amount of time to complete.

When analyzing a new subject, the 3dLift FORTRAN program prompted the user to enter the subject measurements. The program then saved the information to a text file, and referred back to that file when running additional lifting trials for a particular subject. As shown in Table 3.1 for the subject, the parameters $A_1$ through $A_{27}$ were physical measurements. In addition, several parameters were calculated directly from these measurements and are listed as values $A_{28}$ through $A_{35}$ in Table 3.1 and $\psi_1$ through $\psi_{15}$ in Table 3.2. Parameters $A_{28}$ through $A_{35}$ represented segment radii, and were based on assumptions of circular cross sections through the body. These segment radii were used to determine segment moments of inertia (Section 3.3), joint center positions, and segment mass center positions (see Chapter Five). Parameters $\psi_1$ through $\psi_{15}$ were conical measures of the change in radii from the distal to proximal boundaries of the segments. These values were used in finding segment products of inertia (Section 3.3) and in transforming from marker coordinate systems to segment coordinate systems (see Chapter Five). Since the marker and segment coordinate systems corresponded to one another for the upper torso, no conical parameter was needed for this segment.
Table 3.1: Subject Anthropometric Measurements

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Subject (m)</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>A1  Body Mass</td>
<td>77.1 kg</td>
<td>In kg, with subject wearing trial apparel</td>
</tr>
<tr>
<td>A2  Height</td>
<td>1.727</td>
<td>Vertical distance from floor to head vertex</td>
</tr>
<tr>
<td>A3  Foot Length</td>
<td>0.245</td>
<td>Horizontal length from heel to longest toe</td>
</tr>
<tr>
<td>A4  Foot Breadth</td>
<td>0.099</td>
<td>Maximum breadth across metatarsals I-V</td>
</tr>
<tr>
<td>A5  Ankle Height</td>
<td>0.070</td>
<td>Distance from floor to lateral malleolus</td>
</tr>
<tr>
<td>A6  Ankle Circumference</td>
<td>0.240</td>
<td>Ankle circumference at lateral malleolus</td>
</tr>
<tr>
<td>A7  Tibial Height</td>
<td>0.490</td>
<td>Distance from floor to medial tibia</td>
</tr>
<tr>
<td>A8  Tibial Circumference</td>
<td>0.359</td>
<td>Knee circumference at tibia</td>
</tr>
<tr>
<td>A9  Greater Trochanter Height</td>
<td>0.933</td>
<td>Vertical distance from floor to trochanter</td>
</tr>
<tr>
<td>A10 Midthigh Circumference</td>
<td>0.521</td>
<td>At midpoint of tibia and trochanter</td>
</tr>
<tr>
<td>A11 Thigh Circumference</td>
<td>0.619</td>
<td>Circumference just below gluteal furrow</td>
</tr>
<tr>
<td>A12 Superior Iliac Height</td>
<td>1.055</td>
<td>Vertical distance from floor to iliac spines</td>
</tr>
<tr>
<td>A13 Superior Iliac Breadth</td>
<td>0.308</td>
<td>Breadth of torso at superior iliac spines</td>
</tr>
<tr>
<td>A14 Substernale Height</td>
<td>1.228</td>
<td>Vertical distance from floor to substernale</td>
</tr>
<tr>
<td>A15 Substernale Breadth</td>
<td>0.300</td>
<td>Breadth of torso at substernale</td>
</tr>
<tr>
<td>A16 Chest Depth</td>
<td>0.226</td>
<td>Depth of chest at nipples</td>
</tr>
<tr>
<td>A17 Suprasternale Height</td>
<td>1.420</td>
<td>Distance from floor to suprasternale</td>
</tr>
<tr>
<td>A18 Cervicale Height</td>
<td>1.500</td>
<td>Vertical distance from floor to cervicale</td>
</tr>
<tr>
<td>A19 Neck Circumference</td>
<td>0.362</td>
<td>Circumference of neck</td>
</tr>
<tr>
<td>A20 Head Circumference</td>
<td>0.605</td>
<td>Circumference proximal to brow ridges</td>
</tr>
<tr>
<td>A21 Upper Arm Length</td>
<td>0.325</td>
<td>Length from acromion to radiale</td>
</tr>
<tr>
<td>A22 Axillary Circumference</td>
<td>0.329</td>
<td>Circumference of upper arm below armpit</td>
</tr>
<tr>
<td>A23 Elbow Circumference</td>
<td>0.261</td>
<td>Elbow circumference over olecranon</td>
</tr>
<tr>
<td>A24 Lower Arm Length</td>
<td>0.268</td>
<td>Length from radiale to styloin</td>
</tr>
<tr>
<td>A25 Wrist Circumference</td>
<td>0.172</td>
<td>Wrist circumference proximal to styloin</td>
</tr>
<tr>
<td>A26 Fist Length</td>
<td>0.090</td>
<td>Length from styloion to third metacarpal</td>
</tr>
<tr>
<td>A27 Fist Breadth</td>
<td>0.087</td>
<td>Breadth of fist at edge of metacarpals</td>
</tr>
<tr>
<td>A28 Ankle Radius</td>
<td>0.038</td>
<td></td>
</tr>
<tr>
<td>A29 Axillary Radius</td>
<td>0.052</td>
<td></td>
</tr>
<tr>
<td>A30 Elbow Radius</td>
<td>0.042</td>
<td></td>
</tr>
<tr>
<td>A31 Head Radius</td>
<td>0.096</td>
<td></td>
</tr>
<tr>
<td>A32 Neck Radius</td>
<td>0.058</td>
<td></td>
</tr>
<tr>
<td>A33 Tibial Radius</td>
<td>0.057</td>
<td></td>
</tr>
<tr>
<td>A34 Thigh Radius</td>
<td>0.099</td>
<td></td>
</tr>
<tr>
<td>A35 Wrist Radius</td>
<td>0.027</td>
<td></td>
</tr>
</tbody>
</table>
Table 3.2: Segment Conical Parameters

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Segment</th>
<th>Equation (radians)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\psi_1$</td>
<td>Right Foot</td>
<td>$\tan^{-1}((0.5A_4 - A_{28}) / (A_3 - A_{28}))$</td>
</tr>
<tr>
<td>$\psi_2$</td>
<td>Right Calf</td>
<td>$\tan^{-1}((A_{33} - A_{28}) / (A_7 - A_5))$</td>
</tr>
<tr>
<td>$\psi_3$</td>
<td>Right Thigh</td>
<td>$\tan^{-1}((A_{34} - A_{33}) / (A_9 - A_7))$</td>
</tr>
<tr>
<td>$\psi_4$</td>
<td>Left Foot</td>
<td>$\tan^{-1}((0.5A_4 - A_{28}) / (A_3 - A_{28}))$</td>
</tr>
<tr>
<td>$\psi_5$</td>
<td>Left Calf</td>
<td>$\tan^{-1}((A_{33} - A_{28}) / (A_7 - A_5))$</td>
</tr>
<tr>
<td>$\psi_6$</td>
<td>Left Thigh</td>
<td>$\tan^{-1}((A_{34} - A_{33}) / (A_9 - A_7))$</td>
</tr>
<tr>
<td>$\psi_7$</td>
<td>Lower Torso</td>
<td>$\tan^{-1}(0.5A_{16} / (A_{14} - A_9))$</td>
</tr>
<tr>
<td>$\psi_8$</td>
<td>Upper Torso</td>
<td>0.0</td>
</tr>
<tr>
<td>$\psi_9$</td>
<td>Right Upper Arm</td>
<td>$\tan^{-1}((A_{29} - A_{30}) / A_{21})$</td>
</tr>
<tr>
<td>$\psi_{10}$</td>
<td>Right Forearm</td>
<td>$\tan^{-1}((A_{30} - A_{35}) / A_{24})$</td>
</tr>
<tr>
<td>$\psi_{11}$</td>
<td>Right Hand</td>
<td>$\tan^{-1}((0.5A_{27} - A_{35}) / A_{26})$</td>
</tr>
<tr>
<td>$\psi_{12}$</td>
<td>Left Upper Arm</td>
<td>$\tan^{-1}((A_{29} - A_{30}) / A_{21})$</td>
</tr>
<tr>
<td>$\psi_{13}$</td>
<td>Left Forearm</td>
<td>$\tan^{-1}((A_{30} - A_{35}) / A_{24})$</td>
</tr>
<tr>
<td>$\psi_{14}$</td>
<td>Left Hand</td>
<td>$\tan^{-1}((0.5A_{27} - A_{35}) / A_{26})$</td>
</tr>
<tr>
<td>$\psi_{15}$</td>
<td>Head</td>
<td>$\tan^{-1}(0.5A_{16} / (A_2 - A_{17}))$</td>
</tr>
</tbody>
</table>

3.2 Video Marker Placement

In order to determine three-dimensional motions of the lifting subject with the Peak Performance video system, three markers were tracked on each segment. Markers placed directly on the skin and as part of a rigid triad attached to the segment were considered, with each setup having its own advantages and disadvantages (Cappello, 1995). In either case, the markers were positioned to be visible to at least two cameras at all times, if possible. The marker set of each segment was noncollinear. A goal was to attach the markers where the skin movement with respect to the underlying bones was minimal to be consistent with the assumptions of rigid body dynamics. Skin displacements were troublesome if they fell within the frequency range of the overall segment kinematics, since digital filtering could not be used to remove these errors. Using markers attached to rigid triads seemed to eliminate most of the skin movement errors, but since the markers were now located closer together, video errors were magnified. While testing a triad within the lifting motion volume, it was observed that markers should be separated by at least twelve centimeters to avoid excessive angular errors. Finally, skin mounted markers were used on the feet and lower torso because lifting movements appeared awkward while wearing triads attached to these segments.
The initial marker set consisted of thirty-two markers, with every segment defined by three markers. Several markers represented multiple segments, such as the right lateral malleous marker being part of the right foot segment and the right calf segment. After digitizing the sixteen lifting trials, the position data were examined to determine what percentage of time the markers were successfully tracked by the video system. If any marker was tracked less than fifty-five percent over the four trials of each lifting combination, it was considered to have insufficient data and was eliminated from further analysis. Markers placed on the medial elbows showed poor tracking for all lifting combinations since they often faced backwards and also interfered with the lateral elbow markers. Similarly, markers on the medial hands faired poorly during most lifting combinations since they also often faced backward and interfered with the lateral hand markers. Furthermore, a marker placed on the cervicale also suffered from insufficient tracking due to a rearward orientation. During the backlift trials, markers placed on the jaws interfered with markers on the hips and shoulders during the early portion of the motion. This interference occurred in the frontal plane when the back was nearly horizontal in picking up the lifting object from the ground. Finally, markers placed on the superior iliac spines were not properly tracked during the symmetric backlift trials due to interference with the hip markers.

After eliminating the poorly tracked markers, a set of twenty-three markers remained (Table 3.3). The placement of these markers corresponded to the anatomical landmarks discussed earlier in this chapter. In addition, complete marker tracking data from the sixteen lifting trials appear for reference in Appendix B. The remaining markers had an overall tracking percentage of 92.8% for the sixteen lifting trials, which was considered adequate for further signal processing. This tracking percentage will likely improve in future lifting trials if the reduced marker set is used, since less marker-to-marker interference should occur. Unfortunately, removal of markers from the medial elbows and medial hands limited the analysis of the hands, forearms, and upper arms to two degree of freedom segments. Also, the elimination of markers from the jaws limited the analysis of the head to a two degree of freedom segment. This was not a major concern for lifting motions, where the movement of the arms was relatively constrained and the motion of the head was not considered critical.
<table>
<thead>
<tr>
<th>Segment(s)</th>
<th>Marker Placement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot</td>
<td>Right fifth metatarsal (M₁) and left fifth metatarsal (M₇)</td>
</tr>
<tr>
<td>Left Foot</td>
<td>Right first metatarsal (M₂) and left first metatarsal (M₆)</td>
</tr>
<tr>
<td></td>
<td>Right lateral malleolus (M₃)* and left lateral malleolus (M₈)*</td>
</tr>
<tr>
<td>Right Calf</td>
<td>Right lateral malleolus (M₃)* and left lateral malleolus (M₈)*</td>
</tr>
<tr>
<td>Left Calf</td>
<td>Right medial tibiale (M₄)* and left medial tibiale (M₉)*</td>
</tr>
<tr>
<td></td>
<td>Right lateral tibiale (M₅)* and left lateral tibiale (M₁₀)*</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>Right medial tibiale (M₄)* and left medial tibiale (M₉)*</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>Right lateral tibiale (M₅)* and left lateral tibiale (M₁₀)*</td>
</tr>
<tr>
<td></td>
<td>Right greater trochanter (M₁₁)* and left greater trochanter (M₁₂)*</td>
</tr>
<tr>
<td>Lower Torso</td>
<td>Right greater trochanter (M₁₁)*</td>
</tr>
<tr>
<td></td>
<td>Left greater trochanter (M₁₂)*</td>
</tr>
<tr>
<td></td>
<td>Substernale (M₁₃)*</td>
</tr>
<tr>
<td>Upper Torso</td>
<td>Suprasternale (M₁₄)*</td>
</tr>
<tr>
<td></td>
<td>Right acromion (M₁₅)*</td>
</tr>
<tr>
<td></td>
<td>Left acromion (M₁₉)*</td>
</tr>
<tr>
<td>Right Upper Arm</td>
<td>Right acromion (M₁₅)* and left acromion (M₁₉)*</td>
</tr>
<tr>
<td>Left Upper Arm</td>
<td>Right radiale (M₁₆)* and left radiale (M₂₀)*</td>
</tr>
<tr>
<td>Right Forearm</td>
<td>Right radiale (M₁₆)* and left radiale (M₂₀)*</td>
</tr>
<tr>
<td>Left Forearm</td>
<td>Right stylion (M₁₇)* and left stylion (M₂₁)*</td>
</tr>
<tr>
<td>Right Hand</td>
<td>Right stylion (M₁₇)* and left stylion (M₂₁)*</td>
</tr>
<tr>
<td>Left Hand</td>
<td>Right second metacarpal (M₁₈) and left second metacarpal (M₂₂)</td>
</tr>
<tr>
<td>Head</td>
<td>Suprasternale (M₁₄)*</td>
</tr>
<tr>
<td></td>
<td>Top of head or vertex (M₂₃)</td>
</tr>
</tbody>
</table>

* denotes markers shared between segments

The reduced marker set was a major consideration for the division of the torso into upper and lower segments at the T₁₀/T₁₁ intervertebral disk. Use of the T₁₀/T₁₁ disk as a joint center was judged to be most feasible because of its proximity to the substernale marker. Although most disk injuries occur at the L₅/₆₁, this disk was not as easily located since the superior iliac markers were not effectively tracked. Locating the L₅/₆₁ disk from the greater trochanter markers or the substernale marker was viewed as being a potentially large source of error. The T₁₀/T₁₁ disk also had advantages in terms of validation for the 3dLift model, since the T₁₀/T₁₁ approximately split the body and the torso in half. While the focus of this research was the development and initial validation of the 3dLift model, future applications of the model including the L₅/₆₁ disk are considered in Chapter Ten.
3.3 Segment Masses and Moments of Inertia

Segment masses were needed for gravitational forces, for linear inertial terms, and for estimating segment moments of inertia in angular inertial terms. Masses of segments have been commonly predicted as a percentage of a subject's total body mass, based on average values from cadaver or scanning studies. Differences in body shapes have also been considered when predicting segment masses by multiplying the segment volume by a uniform segment density. Alternatively, regression equations based on cadaver and scanning studies have combined segment mass and geometry factors to account for both shape and nonuniform densities. Segment masses have also been estimated on individual subjects using experimental techniques such as computer tomography, but cost could be a limiting factor with such methods. It should be noted that the segment masses were based on predominantly adult male subjects and further adjustments are in order when studying females or children. The percentages of total body mass used to estimate individual segment masses were in general agreement among previous researchers and thus this method was used. In addition, the lifting subject was male and was within the range of masses for the subjects in the published studies. Segment masses as a percentage of total body mass are listed in Table 3.4 along with subject values as calculated in the 3dLift biomechanical model. Additional background on how segment mass percentages were calculated in Table 3.4 appears in Appendix C.

Referring to Table 3.4, the foot and lower leg mass percents were based on the average of predictions by Clauser et al. (1969), Zatsiorsky and Seluyanov (1983), and Clarys and Marfell-Jones (1994). Averaging was deemed appropriate since the mass percents varied less than one percent from one another between the three research groups for these segments. Similarly, the upper arm, forearm, hand, and head mass percents were also based on the average of results from these three sources. Again, the estimated mass percents varied less than one percent when comparing the three studies for these segments. The lower and middle torso mass of Zatsiorsky and Seluyanov (1983) was averaged with the lower torso to tenth thoracic vertebrae mass of Pearsall et al. (1996). These studies were used because they corresponded to the torso definitions of the 3dLift model, and they varied by less than one percent. In addition, the upper torso mass of Zatsiorsky and Seluyanov (1983) was also
Table 3.4: Segment Masses as a Percentage of Total Body Mass

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mass Percent</th>
<th>Subject (kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>m_1 Right Foot</td>
<td>1.45</td>
<td>1.1</td>
</tr>
<tr>
<td>m_2 Right Calf</td>
<td>4.1</td>
<td>3.2</td>
</tr>
<tr>
<td>m_3 Right Thigh</td>
<td>14.0</td>
<td>10.8</td>
</tr>
<tr>
<td>m_4 Left Foot</td>
<td>1.45</td>
<td>1.1</td>
</tr>
<tr>
<td>m_5 Left Calf</td>
<td>4.1</td>
<td>3.2</td>
</tr>
<tr>
<td>m_6 Left Thigh</td>
<td>14.0</td>
<td>10.8</td>
</tr>
<tr>
<td>m_7 Lower Torso</td>
<td>28.0</td>
<td>21.6</td>
</tr>
<tr>
<td>m_8 Upper Torso</td>
<td>16.0</td>
<td>12.3</td>
</tr>
<tr>
<td>m_9 Right Upper Arm</td>
<td>2.7</td>
<td>2.1</td>
</tr>
<tr>
<td>m_10 Right Forearm</td>
<td>1.51</td>
<td>1.2</td>
</tr>
<tr>
<td>m_11 Right Hand</td>
<td>0.62</td>
<td>0.5</td>
</tr>
<tr>
<td>m_12 Left Upper Arm</td>
<td>2.7</td>
<td>2.1</td>
</tr>
<tr>
<td>m_13 Left Forearm</td>
<td>1.51</td>
<td>1.2</td>
</tr>
<tr>
<td>m_14 Left Hand</td>
<td>0.62</td>
<td>0.5</td>
</tr>
<tr>
<td>m_15 Head</td>
<td>7.2</td>
<td>5.6</td>
</tr>
</tbody>
</table>

compared to the first to tenth thoracic vertebrae mass of Pearsall et al. (1996). There was only about a three percent difference in this case, but the percentage of Zatsiorsky and Seluyanov (1983) was used when later considering the range of percentages for the upper legs. Finally, the upper legs show the greatest variation between studies, and their mass percent was estimated by simply dividing by remaining total body mass percent by two. The results then fell within the range of values predicted by Clauser et al. (1969), Zatsiorsky and Seluyanov (1983), and Clarys and Marfell-Jones (1994).

The moments of inertia and products of inertia for each segment of the 3dLiit model were needed to determine the rotational inertial terms in the equations of motion. To account for differences in subject body types, results from cadaver or scanning studies have been analyzed to find the radius of gyration as a percentage of segment length. Another technique represented the segments as simplified geometries that have a known moment of inertia and product of inertia formulas, to which the estimated segment mass was applied. Again referencing cadaver or scanning studies, segment moments of inertia have also been predicted using regression equations of various levels of complexity. Unfortunately, when compared to the segment masses, there was less agreement among researchers about segment moments of inertia. Due to this variance, moments of inertia were predicted using segment
geometrical parameters contained in equations that were adjusted to compromise between literature values. Estimations of segment products of inertia were described infrequently in previous research, and thus these values were estimated strictly on geometrical considerations (Kane and Levinson, 1985). The equations used by the 3dLift biomechanical model for predicting segment moments of inertia and products of inertia appear in Tables 3.5 and 3.6. Appendix C contains additional information underlying the moment of inertia equations of Tables 3.5 and 3.6.

Tables 3.5-3.6 refer to the anthropometric parameters listed in Tables 3.1-3.2, and the $m_n$ variables represent the mass of segment $n$ determined using Table 3.4. The i-axes were predominantly in the anterior posterior direction, the j-axes in the medial-lateral direction, and the k-axes in the vertical direction as the subject stood in the anatomical position (see Chapter Five). To determine the equations, moment of inertia values for each segment were determined from multiple literature sources and averaged, adjusting the segment definitions if necessary (de Leva, 1996a). The average result was then used to scale the equations in the form of Vaughan et al. (1992) for the lower extremity and Hanavan (1964) for the remaining segments.

Foot moments of inertia were determined by averaging the regression equations of Zatsiorsky and Seluyanov (1983) with the regression equations of Vaughan et al. (1992). Products of inertia were predicted for the feet by assuming a trapezoidal cross-section along the i-j plane, symmetry along the j-k plane, and a right triangle along the i-k plane. Next, moments of inertia for the calf and thigh were developed using the model of Hanavan (1964), the equations of Zatsiorsky and Seluyanov (1983), and the adjustments of de Leva (1996a). Calf and thigh products of inertia were estimated by assuming truncated cones that were symmetrical about their i-j planes and formed similar trapezoids along their j-k and i-k planes. To complete the lower body segments, the lower torso moments of inertia were an average of the model of Hanavan (1964) and the computer tomography study of Pearsall et al. (1996). The lower torso was modeled as a right elliptical cylinder, thus only the shift of the mass center from the geometrical center appeared in the products of inertia.

As with the lower torso, the upper torso moments of inertia were an average of the model of Hanavan (1964) and the computerized tomography study of Pearsall et al. (1996). The
Table 3.5: Lower Body Segment Moments of Inertia and Products of Inertia

<table>
<thead>
<tr>
<th>Segment (n)</th>
<th>Moments and Products of Inertia (kg·m²)</th>
<th>Subject</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot (1)</td>
<td></td>
<td>1.33×10⁻³</td>
</tr>
<tr>
<td>I_{ii} = 0.085m_n(\bar{A}_4^2 + \bar{A}_5^2) - 0.0007</td>
<td></td>
<td></td>
</tr>
<tr>
<td>I_{jj} = 0.0165m_n(3\bar{A}_2^2 + 4\bar{A}_5^2) + 0.00023</td>
<td>4.04×10⁻³</td>
<td></td>
</tr>
<tr>
<td>I_{kk} = 0.0148m_n(3\bar{A}_3^2 + 4\bar{A}_4^2) + 0.00068</td>
<td>4.42×10⁻³</td>
<td></td>
</tr>
<tr>
<td>I_{ij} = I_{ji} = m_n\bar{A}_{28} \sin(2\psi_n) / 6</td>
<td>2.90×10⁻⁵</td>
<td></td>
</tr>
<tr>
<td>I_{ik} = I_{ki} = m_nA_3(A_3 - A_{28}) / 36</td>
<td>4.59×10⁻⁴</td>
<td></td>
</tr>
<tr>
<td>I_{jk} = I_{kj} = 0</td>
<td>0.0</td>
<td></td>
</tr>
</tbody>
</table>

| Left Foot (4)       |                                                                                                          |          |
| I_{ii} = 0.0165m_n(3\bar{A}_2^2 + 4\bar{A}_5^2) + 0.00023                                                                 | 4.53×10⁻²|
| I_{jj} = 0.0148m_n(3\bar{A}_3^2 + 4\bar{A}_4^2) + 0.00068                                                                 | 4.35×10⁻²|
| I_{kk} = 0.0100m_n\bar{A}_8^2 + 0.000119                                                                      | 4.19×10⁻³|
| I_{ij} = I_{ji} = 0                                                                                          | 0.0      |
| I_{ik} = I_{ki} = m_n\bar{A}_{28} \sin(2\psi_n) / 6                                                            | 6.92×10⁻⁵|
| I_{jk} = I_{kj} = m_n\bar{A}_{28} \sin(2\psi_n) / 6                                                            | 6.92×10⁻⁵|

| Right Calf (2)      |                                                                                                          | 4.53×10⁻²|
| I_{ii} = 0.075m_n(\bar{A}_7 - \bar{A}_5)^2 + 0.076A_8^2 + 0.00111                                              |          |
| I_{jj} = 0.067m_n(\bar{A}_7 - \bar{A}_5)^2 + 0.076A_8^2 + 0.0041                                                | 4.35×10⁻²|
| I_{kk} = 0.0100m_n\bar{A}_8^2 + 0.000119                                                                      | 4.19×10⁻³|
| I_{ij} = I_{ji} = 0                                                                                          | 0.0      |
| I_{ik} = I_{ki} = m_n\bar{A}_{28} \sin(2\psi_n) / 6                                                            | 6.92×10⁻⁵|
| I_{jk} = I_{kj} = m_n\bar{A}_{28} \sin(2\psi_n) / 6                                                            | 6.92×10⁻⁵|

| Left Calf (5)       |                                                                                                          |          |
| I_{ii} = 0.070m_n(\bar{A}_7 - \bar{A}_5)^2 + 0.076A_8^2 + 0.0161                                              | 1.80×10⁻¹|
| I_{jj} = 0.070m_n(\bar{A}_7 - \bar{A}_5)^2 + 0.076A_8^2 + 0.0149                                                | 1.79×10⁻¹|
| I_{kk} = 0.0108m_n\bar{A}_{10} + 0.0031                                                                           | 3.47×10⁻²|
| I_{ij} = I_{ji} = 0                                                                                           | 0.0      |
| I_{ik} = I_{ki} = m_n\bar{A}_{32} \sin(2\psi_n) / 6                                                             | 1.09×10⁻³|
| I_{jk} = I_{kj} = m_n\bar{A}_{32} \sin(2\psi_n) / 6                                                             | 1.09×10⁻³|

| Right Thigh (3)     |                                                                                                          | 1.80×10⁻¹|
| I_{ii} = 0.092m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{13}^2                                                     |          |
| I_{jj} = 0.098m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{16}^2                                                     | 2.65×10⁻¹|
| I_{kk} = 0.26m_n(0.25\bar{A}_{13}^2 + \bar{A}_{16}^2)                                                          | 2.05×10⁻¹|
| I_{ij} = I_{ji} = I_{ik} = I_{kj} = 0                                                                           | 0.0      |
| I_{ik} = I_{ki} = -0.000129m_n\bar{A}_{16}(\bar{A}_4 - \bar{A}_9)                                              | -1.86×10⁻³|
| I_{jk} = I_{kj} =                                                                                               |          |

| Left Thigh (6)      |                                                                                                          |          |
| I_{ii} = 0.092m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{13}^2                                                     |          |
| I_{jj} = 0.098m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{16}^2                                                     | 2.65×10⁻¹|
| I_{kk} = 0.26m_n(0.25\bar{A}_{13}^2 + \bar{A}_{16}^2)                                                          | 2.05×10⁻¹|
| I_{ij} = I_{ji} = I_{ik} = I_{kj} = 0                                                                           | 0.0      |
| I_{ik} = I_{ki} = -0.000129m_n\bar{A}_{16}(\bar{A}_4 - \bar{A}_9)                                              | -1.86×10⁻³|
| I_{jk} = I_{kj} =                                                                                               |          |

| Lower Torso (7)     |                                                                                                          |          |
| I_{ii} = 0.092m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{13}^2                                                     |          |
| I_{jj} = 0.098m_n(\bar{A}_4 - \bar{A}_9)^2 + 0.75A_{16}^2                                                     | 2.65×10⁻¹|
| I_{kk} = 0.26m_n(0.25\bar{A}_{13}^2 + \bar{A}_{16}^2)                                                          | 2.05×10⁻¹|
| I_{ij} = I_{ji} = I_{ik} = I_{kj} = 0                                                                           | 0.0      |
| I_{ik} = I_{ki} = -0.000129m_n\bar{A}_{16}(\bar{A}_4 - \bar{A}_9)                                              | -1.86×10⁻³|
| I_{jk} = I_{kj} =                                                                                               |          |

The upper torso was assumed to be a right elliptical cylinder with the products of inertia arising from the shift of the segment geometrical center to the mass center. Next, the moments of inertia of the upper arms and forearms were an average of the model of Hanavan (1964) and the equations of Zatsiorsky and Seluyanov (1983). Similar to the legs, the upper arms and
Table 3.6: Upper Body Segment Moments of Inertia and Products of Inertia

<table>
<thead>
<tr>
<th>Segment (n)</th>
<th>Moments and Products of Inertia (kg·m²)</th>
<th>Subject</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper Torso (8)</td>
<td>$I_{ii} = 0.070m_n \left(0.75A_{15}^2 + A_{18} - A_{14}\right)^2$</td>
<td>1.22x10⁻¹</td>
</tr>
<tr>
<td></td>
<td>$I_{jj} = 0.062m_n \left(0.75A_{16}^2 + A_{18} - A_{14}\right)^2$</td>
<td>8.59x10⁻²</td>
</tr>
<tr>
<td></td>
<td>$I_{kk} = 0.26m_n \left(0.75A_{15}^2 + 0.25A_{16}^2\right)$</td>
<td>1.13x10⁻¹</td>
</tr>
<tr>
<td></td>
<td>$I_{ij} = I_{ji} = I_{jk} = I_{kj} = 0$</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>$I_{ik} = I_{ki} = -0.000106m_nA_{16} (A_{18} - A_{14})$</td>
<td>-8.04x10⁻⁴</td>
</tr>
<tr>
<td>Right Upper Arm (9)</td>
<td>$I_{ii} = 0.080m_n \left(A_{21}^2 + 0.076A_{22}^2\right)$</td>
<td>1.90x10⁻²</td>
</tr>
<tr>
<td>Left Upper Arm (12)</td>
<td>$I_{jj} = 0.076m_n \left(A_{21}^2 + 0.076A_{22}^2\right)$</td>
<td>1.80x10⁻²</td>
</tr>
<tr>
<td></td>
<td>$I_{kk} = 0.021m_nA_{22}^2$</td>
<td>4.73x10⁻³</td>
</tr>
<tr>
<td></td>
<td>$I_{ij} = I_{ji} = 0$</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>$I_{ik} = I_{ki} = m_nA_{30}^2 \sin(2\psi_n)/6$</td>
<td>3.98x10⁻⁵</td>
</tr>
<tr>
<td></td>
<td>$I_{jk} = I_{kj} = m_nA_{30}^2 \sin(2\psi_n)/6$</td>
<td>3.98x10⁻⁵</td>
</tr>
<tr>
<td>Right Forearm (10)</td>
<td>$I_{ii} = 0.082m_n \left(A_{24}^2 + 0.076A_{23}^2\right)$</td>
<td>7.35x10⁻³</td>
</tr>
<tr>
<td>Left Forearm (13)</td>
<td>$I_{jj} = 0.076m_n \left(A_{24}^2 + 0.076A_{23}^2\right)$</td>
<td>6.81x10⁻³</td>
</tr>
<tr>
<td></td>
<td>$I_{kk} = 0.0104m_nA_{23}^2$</td>
<td>8.25x10⁻⁴</td>
</tr>
<tr>
<td></td>
<td>$I_{ij} = I_{ji} = 0$</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>$I_{ik} = I_{ki} = m_nA_{35}^2 \sin(2\psi_n)/6$</td>
<td>1.53x10⁻⁵</td>
</tr>
<tr>
<td></td>
<td>$I_{jk} = I_{kj} = m_nA_{35}^2 \sin(2\psi_n)/6$</td>
<td>1.53x10⁻⁵</td>
</tr>
<tr>
<td>Right Hand (11)</td>
<td>$I_{ii} = 0.159m_nA_{27}^2$</td>
<td>5.75x10⁻⁴</td>
</tr>
<tr>
<td>Left Hand (14)</td>
<td>$I_{jj} = 0.159m_nA_{27}^2$</td>
<td>5.75x10⁻⁴</td>
</tr>
<tr>
<td></td>
<td>$I_{kk} = 0.159m_nA_{27}^2$</td>
<td>5.75x10⁻⁴</td>
</tr>
<tr>
<td></td>
<td>$I_{ij} = I_{ji} = I_{ik} = I_{ki} = I_{jk} = I_{kj} = 0$</td>
<td>0.0</td>
</tr>
<tr>
<td>Head (15)</td>
<td>$I_{ii} = 0.21m_n \left(0.25(A_2 - A_{18})^2 + A_{31}^2\right)$</td>
<td>2.58x10⁻²</td>
</tr>
<tr>
<td></td>
<td>$I_{jj} = 0.23m_n \left(0.25(A_2 - A_{18})^2 + A_{31}^2\right)$</td>
<td>2.83x10⁻²</td>
</tr>
<tr>
<td></td>
<td>$I_{kk} = 0.38m_nA_{31}^2$</td>
<td>1.96x10⁻²</td>
</tr>
<tr>
<td></td>
<td>$I_{ij} = I_{ji} = I_{ik} = I_{ki} = I_{jk} = I_{kj} = 0$</td>
<td>0.0</td>
</tr>
</tbody>
</table>
forearms were assumed to be truncated cones with symmetrical i-j planes and trapezoidal i-k and j-k planes to calculate the products of inertia. For the hands, the k-axis moment of inertia equations of Zatsiorsky and Seluyanov (1983) were used to scale the spherical fist model of Hanavan (1964). Since the fist model is spherical with a mass center displacement only along the k-axis, there is symmetry about all planes and the products of inertia were zero. Finally, the moments of inertia for the head were estimated by again using the model of Hanavan (1964) and the equations of Zatsiorsky and Seluyanov (1983). The head was assumed to be a right circular ellipsoid with symmetry about all planes and a mass center displacement only along the k-axis, and thus the products of inertia were zero.

This chapter provided the first steps in the development of the 3dLift biomechanical model by quantifying anthropometric parameters of the lifting subject. In order to achieve this objective, a set of physical measurements was listed that were performed on the subject. Next, the placement of reflective markers on anatomical landmarks was considered for tracking by video equipment. Finally, estimations of segment mass, moments of inertia, and products of inertia were determined for later use in the equations of motion. In the next chapter, the raw data from the video tracking equipment are prepared for further analysis through signal processing. Methods will be discussed for dealing with missing data points and for removing erroneous noise from the data.
When examining the raw (unprocessed) video and force platform output, two classes of errors were found: missing data and noisy data. Signal processing by the 3dLift biomechanical model was therefore required to remove as much of the erroneous data as possible before further analyzing the data. Missing data were only a concern with the video measurements and occurred when markers were visible by less than two out of four cameras. This situation arose when body segments or the lifting object obscured markers and when two markers passed within close proximity of one another. Noisy data included signals measured from sources other than those intended by the experimental equipment. Force platform noise may have originated from other electrical devices, surrounding vibrations, and cross talk between the strain gauges. Video noise may have been caused by reflections in the laboratory, marker flickering between adjacent camera pixels, and errors inherent in the solution of the coordinate reconstruction. Extrapolation was used when data were missing at the beginning or end of the lifting trial, interpolation was used when intermediate data were missing during the lifting trial, and digital filtering was used to reduce signal noise.

Prior to the lifting trials, video data were collected while the subject stood motionless in the anatomical position. Marker coordinates were automatically digitized. As listed in Table 3.3, the 3dLift program then calculated the length between the markers associated with each segment. Several additional lengths of interest were found, such as the left greater trochanter to the suprasternal and the mid-acromions to the substermale, suprastemale, and the vertex. These lengths were used to estimate missing data points for a troublesome class of extrapolation problems (Section 4.1). It was also of interest to divide the lifting trials into time periods corresponding to pre-lift, lifting, and post-lift.

The pre-lift phase began at the first data point collected and ended at initiation of the lifting motion. Furthermore, the lifting phase started at lift initiation and ended with lift conclusion and the post-lift phase began at lift conclusion and ended with the last data point collected. The 3dLift program automatically determined the lift phases by checking the vertical position of the vertex (top of head) marker. The initiation of the lift was found by detecting the beginning of a steady increase in the vertical position of the vertex, and the
conclusion of the lift was found by detecting the end of this increase. More specifically, the
lift initiation was defined to be at least ten consecutive increases in the vertical position of the
vertex marker. Such a scheme was effective in determining the initiation of movement
related to lifting, while ignoring oscillations of the head that might occur prior to the lift.
Vertical force platform ground reactions were also considered for this purpose, which
worked well for the initiation of the lift, but was inconsistent in finding the conclusion of the
lift.

4.1 Extrapolation and Interpolation

The 3dLift program used extrapolation methods to estimate missing video data at the start
of the pre-lift and the end of the post-lift time periods. It was assumed that little motion
occurred during these portions of the pre-lift and post-lift time periods. If partial video data
existed in these time periods, then a pre-lift and/or post-lift average marker position was
calculated. For these cases, the missing pre-lift marker positions were filled with average
pre-lift positions, and the missing post-lift marker positions were filled with average post-lift
positions. These steps were extended for cases of remaining missing marker data for the feet
throughout the entire lifting trial. In such cases, an average marker position was determined
over the entire trial, and the missing marker data were replaced with these average positions.
As before, this approach was warranted since the feet remained approximately stationary on
the force platforms during the lifting trials. If movement of the feet was required during the
motion of interest, then alternative methods such as those described below would have to be
used.

A second, more difficult case where extrapolation was needed occurred when a marker
was obscured for the entire pre-lift time period. These obscured marker positions were
estimated using visible surrounding markers and the approximate lengths between markers
determined from the anatomical position data. The obscured marker positions were
calculated using a weighted average of an initial guess and a corrected position:

\[ M = \frac{(M_{SD} + 3M_{IP})}{4}, \]  

(4.1)

where \( M \) was the estimated obscured marker position, \( M_{SD} \) was the corrected marker
position, and \( M_{IP} \) was the initial marker position guess. Corrected marker positions were
found using the method of steepest descent for adjusting the initial marker positions to better match the known lengths between markers:

\[ f(M_m) = \left( |M_n - M_m|^2 - L_{mn}^2 \right)^2 + \left( |M_o - M_m|^2 - L_{mo}^2 \right)^2 + \left( |M_p - M_m|^2 - L_{mp}^2 \right)^2, \]

where \( f \) was the minimization function, \( M_m \) was the obscured marker, \( M_n \) to \( M_p \) were the visible markers, and \( L_{mn} \) to \( L_{mp} \) were the known lengths between markers. The method of steepest descent was used to find a local minimum by using the gradient to determine a direction from the initial guess that decreased the length function value (Burden and Faires, 1989). A weighted average was used since there were errors in the corrected values due to skin displacements of the markers between the anatomical and pre-lift body positions. Other numerical analyses such as quasi-Newton methods were tested, but the problem tended to be ill-conditioned, probably also due to skin displacements. A potential improvement to the method could be to digitize multiple subject positions when calculating marker lengths. Subroutine Descent and Function Sumsquares from the 3dLift program implemented the steepest descent method and appears in Appendix D.

Table 4.1 displays the options for surrounding visible markers used as reference points and the initial guesses of the marker positions. Multiple solution options were available for several of the markers. These are listed in the order they were checked for applicability by the 3dLift program. Initial guesses in the i-, j-, and k-directions were required, and the visible markers whose coordinates best approximated the obscured marker in each direction were utilized. For the symmetric leglift trials, the left hip marker was most likely to be obscured prior to the lift due to the orientation of the cameras and the positioning of the arms. In the asymmetric leglift trials, the left lateral knee marker was most likely to be obscured during prelift. For both the symmetric and asymmetric backlift trials, the lower sternum marker was most likely to be obscured at prelift due to the orientation of the upper body. Since both the medial and lateral right knee markers were obscured prior to lifting in trial seven (asymmetric leglift), this trial was dropped from further analysis. The remaining fifteen lifting trials all fell within the surrounding visible marker guidelines listed in Table 4.1.
Table 4.1: Extrapolation of Obscured Markers

<table>
<thead>
<tr>
<th>Obscured Marker</th>
<th>Option</th>
<th>Visible Markers</th>
<th>Initial Guess, i</th>
<th>Initial Guess, j</th>
<th>Initial Guess, k</th>
</tr>
</thead>
<tbody>
<tr>
<td>M12</td>
<td>1</td>
<td>M11, M9, M10</td>
<td>M11</td>
<td>M10</td>
<td>M11</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>M11, M10, M13</td>
<td>M11</td>
<td>M10</td>
<td>M11</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>M11, M9, M13</td>
<td>M11</td>
<td>M9</td>
<td>M11</td>
</tr>
<tr>
<td>M5</td>
<td></td>
<td>M3, M4, M11</td>
<td>M4</td>
<td>M3</td>
<td>M4</td>
</tr>
<tr>
<td>M10</td>
<td></td>
<td>M8, M9, M12</td>
<td>M9</td>
<td>M8</td>
<td>M9</td>
</tr>
<tr>
<td>M4</td>
<td></td>
<td>M3, M5, M11</td>
<td>M5</td>
<td>M3</td>
<td>M5</td>
</tr>
<tr>
<td>M9</td>
<td></td>
<td>M8, M10, M12</td>
<td>M10</td>
<td>M8</td>
<td>M10</td>
</tr>
<tr>
<td>M13</td>
<td>1</td>
<td>M11, M15, M14</td>
<td>(M11 + M14)/2</td>
<td>M14</td>
<td>(M11 + M14)/2</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>M11, M15, M12</td>
<td>(M15 + M12)/2</td>
<td>M15</td>
<td>(M15 + M12)/2</td>
</tr>
<tr>
<td>M14</td>
<td>1</td>
<td>M13, M15, M19</td>
<td>(M13 + M15)/2</td>
<td>M13</td>
<td>(M13 + M15)/2</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>M13, M15, M11</td>
<td>(M13 + M15)/2</td>
<td>M13</td>
<td>(M13 + M15)/2</td>
</tr>
<tr>
<td>M15</td>
<td></td>
<td>M14, M15, M19</td>
<td>M19</td>
<td>M16</td>
<td>M19</td>
</tr>
<tr>
<td>M19</td>
<td></td>
<td>M14, M15, M20</td>
<td>M15</td>
<td>M20</td>
<td>M15</td>
</tr>
</tbody>
</table>

Markers are listed in Table 3.3

Interpolation methods were utilized when marker position data were obscured within intermediate points of the lifting trial. A typical gap in the position data was fifteen data points or one-eighth of a second. Although the steepest descent method could have been used again, it suffered from displacement errors as the skin moved with respect to the rigid bones. As an alternative, since it was now guaranteed that missing data were preceded and followed by valid data, interpolation provided a simpler and more accurate solution. For an initial step, the 3dLift program searched for the three data points occurring just prior to and after the missing data. Next, three estimations using linear interpolation were performed utilizing these six visible marker positions:

\[
\begin{align*}
M_{na} &= \frac{(t_n - t_{n-3})(M_{n+1} - M_{n-3})}{t_{n+1} - t_{n-3}} + M_{n-3}, \\
M_{nb} &= \frac{(t_n - t_{n-2})(M_{n+2} - M_{n-2})}{t_{n+2} - t_{n-2}} + M_{n-2}, \\
M_{nc} &= \frac{(t_n - t_{n-1})(M_{n+3} - M_{n-1})}{t_{n+3} - t_{n-1}} + M_{n-1},
\end{align*}
\]

where \( M_{na} \) to \( M_{nc} \) were obscured marker interpolations, \( M_{n-3} \) to \( M_{n+3} \) were visible marker data, \( t_n \) was interpolated marker time, and \( t_{n-3} \) to \( t_{n+3} \) were visible marker times.
Multiple interpolations were performed to reduce the effects of signal noise on the estimated points since the position data had yet to be filtered. Three data points were used as a compromise between noise reduction with multiple points and using as few points as necessary to allow for rapid changes in position. The estimated marker position was then simply set to the average of the three linear interpolations:

\[ M_n = \frac{(M_{na} + M_{nb} + M_{nc})}{3}, \]

where \( M_n \) is the estimated marker position. Cubic spline interpolation was also tested, but resulted in undesirable oscillations being introduced into the data.

4.2 Digital Filtering of Experimental Data

When collecting lifting trials with the video system and the force platforms, the measured signals included a certain amount of noise or data unrelated to the motion of interest. Examining the content of the signal and filtering out the unwanted frequencies often eliminated much of this noise. In accordance with the Nyquist sampling theorem, the maximum frequency that can be accurately measured is one-half the sampling frequency, known as the Nyquist frequency. The frequencies above the Nyquist frequency are instead undersampled, resulting in aliasing. When aliasing occurs, signals above the Nyquist frequency are reflected back onto the lower frequencies and result in erroneous signals in the time domain. Therefore, sampling frequencies should be selected that are twice the frequency of both the signals of interest and the expected sources of noise.

When measuring lifting motions, the frequencies of interest were expected to fall below 10 Hz, thus a sampling frequency of at least 20 Hz was required. In practice, sampling frequencies four to five times the highest expected frequency attributed to human movement have been used (Baker, 1994). As an additional consideration, a sampling frequency of 120 Hz was chosen since it was twice that of the 60 Hz alternating current powering surrounding electrical devices. This ensured that most of the electrical sources of noise could be attenuated through digital filtering. Another factor in this decision was aiding in the synchronization of the two sets of data by making the video and force platform sampling frequency equal to one another. Therefore, both the force platforms and the video cameras were set to sample data at 120 Hz.
A popular method for reducing noise in motion analysis has been the use of cubic and quintic splines that fit raw data with groups of low order polynomial functions. Another common method has been the inverse Fast Fourier Transform, which involves conversions between the time and frequency domains (Kahaner et al., 1989). When skin movement induced errors are present, a least-squares optimization scheme that minimizes changes in lengths between the markers has been also been suggested (Cappozzo and Cappello, 1997). The data associated with the lifting motions was expected to have relatively high magnitudes and be restricted to low frequencies. In contrast, the signal noise was expected to have lower magnitudes and to occur at higher frequencies in the spectrum. These signal characteristics made it possible to use a low-pass digital filter to eliminate high frequency noise from the data (Winter, 1990). A low-pass filter leaves data below a selected cutoff frequency (Section 4.3) unchanged and attenuates data above the cutoff, with a transitional area (rolloff) that partially attenuates the signal. Specifically, a fourth-order Butterworth low-pass filter was chosen because it would give equivalent results to a quintic spline if both were used properly (Woltring, 1995).

The Butterworth filter was recursive in nature, which means that it used both raw data and previously filtered data to eliminate noise. Recursive or infinite impulse response filters had an advantage over non-recursive or finite impulse response filters, in that they could be designed with sharper rolloffs. This resulted in a narrowing of the transitional zone between the unchanged low frequency signals and the rejected high frequency signals. Second-order recursive filters had the following format:

\[ Y(nT) = a_0X(nT) + a_1X(nT - T) + a_2(nT - 2T) + b_1Y(nT - T) + b_2Y(nT - 2T), \quad (4.5) \]

where \( Y \) was filtered data, \( X \) was raw data, \( n \) was the data point, \( T \) was the sampling period, and \( a_0 \) to \( b_2 \) were filter coefficients. For a Butterworth filter, the coefficients were dependent on the ratio of the sampling frequency to the cutoff frequency and were calculated using (Winter, 1990):

\[
\omega_c = \tan \left( \frac{\pi f_c}{f_s} \right), \quad K_1 = \sqrt{2}\omega_c, \quad (4.6a, b)
\]

\[
K_2 = \omega_c^2, \quad a_0 = \frac{K_2}{(1 + K_1 + K_2)}, \quad (4.6c, d)
\]
\[ a_1 = 2a_0, \quad a_2 = a_0, \quad (4.6e, f) \]
\[ K_3 = \frac{2a_0}{K_2}, \quad b_1 = -2a_0 + K_3, \quad (4.6g, h) \]
\[ b_2 = 1 - 2a_0 - K_3, \quad (4.6i) \]

where \( f_c \) was the cutoff frequency and \( f_s \) was the sampling frequency. The sampling frequencies were 120 Hz for both the force platforms and video cameras. The selection of cutoff frequencies is discussed in Section 4.3.

Since the Butterworth filter was a nonsymmetric, recursive filter, a phase lag was introduced into the data when it was used. Filtering the data twice, once forward and once backwards, with respect to time eliminated this phase lag. Not only did this second filtering pass cancel out the time distortions, but it also further sharpened the rolloff, resulting in a fourth order, zero-phase-shift Butterworth filter. Since the filter was passed through twice, the transitional region became steeper and the cutoff frequency had to be adjusted before determining the Butterworth coefficients (Winter, 1990). The adjustment was dependent on the cutoff to sampling frequency ratio and ranged from 1.0 at the Nyquist frequency to an asymptotic value of 0.802 at continuous sampling. After converting the cutoff frequency from Hertz to radians per second, the adjusted cutoff frequency was calculated as follows (Dowling, 1982):

\[ \omega_c = 2\pi f_c, \quad \Omega_c = \tan\left(\frac{\omega_c}{2f_s}\right), \quad (4.9a) \]
\[ \Omega_s = \frac{\Omega_c}{(\sqrt{2} - 1)^{0.25}}, \quad \omega_s = 2f_s \tan^{-1} \Omega_s, \quad (4.9b) \]
\[ f_s = \frac{\omega_s}{2\pi}, \quad (4.9c) \]

where \( \omega \) terms were angular frequencies, \( \Omega \) terms were transformed frequencies, and \( f_s \) was the adjusted cutoff frequency.

When reducing noise for a data point, the Butterworth filter used two previously analyzed data points (Equation 4.5). To initialize the filter, the first data point remained unchanged,
and the second and third points were run through a weighted three-point moving average to eliminate high frequency noise:

\begin{align}
    Y(1) &= X(1), \\
    Y(2) &= \frac{X(1)+2X(2)+X(3)}{4}, \\
    Y(3) &= \frac{X(2)+2X(3)+X(4)}{4}, \\
    Y(4) &= a_0 X(4) + a_1 X(3) + a_2 X(2) + b_1 Y(3) + b_2 Y(2),
\end{align}

where \( Y \) was filtered data, \( X \) was raw data, and \( a_0 \) to \( b_2 \) were Butterworth filter coefficients (Equation 2.6). The remaining raw data, points five to the last point sampled, were also run through the Butterworth filter. When running the data through the Butterworth filter a second time backwards, the same system was used in reverse. Although the first three and last three data points were expected to contain a higher level of noise, this was not of concern since these points were separated in time from the lifting motion. As an example of the signal processing methods used in this research, the vertical position of the left hip marker for trial one is shown in Figure 4.1. This trial was a symmetric leglift, and in addition to digital filtering, the poorly tracked left hip marker data required the steepest descent extrapolation and interpolation. The subroutine Butterworth from the 3dLift FORTRAN program appears for reference in Appendix D.

### 4.3 Selection of the Cutoff Frequency

When signal processing has been used to reduce noise in data, the characteristics of the signal that should remain and those that should be removed have been prescribed by various means. Spline programs often have required a smoothing factor to be assigned that indicated how sharply the data could vary over time. The smoothing factor for cubic and quintic splines is related to the cutoff frequency used with the Butterworth digital filter (van den Bogert, 1995):

\begin{align}
    \alpha &= f_s \left( \frac{2\pi f_c}{k_f} \right)^{-2m}, \\
    k_f &= (\sqrt{2} - 1)^{0.5m}, \quad m = \left( s_0 + \frac{1}{2} \right),
\end{align}
Figure 4.1: Signal Processing for Left Hip Marker, Trial One (Symmetric Leglift)
where $\alpha$ was the smoothing factor, $f_s$ was the sampling frequency, $f_c$ was the cutoff frequency, $k_f$ was the scaling factor, $m$ was the order adjustment, and $s_0$ was the spline order. One optimization approach for choosing the smoothing factor was called generalized cross validation and was part of a spline package called GCVSPL (Woltring, 1995). Another optimization algorithm called LAMBDA was claimed to improve on cross validation through use of an autoregressive derivative assessment (D'Amico and Ferrigno, 1992). Alternatively, the power spectrum calculated using the Fast Fourier Transform has been plotted to indicate the strength of the signal at different frequencies. The higher frequencies that were separated from the low frequency components of the signal could then be removed using the inverse Fast Fourier Transform. Regardless of the method employed, the goal was to eliminate as much of the noise as possible with minimum attenuation of the actual signal.

When using the Butterworth digital filter, a cutoff was chosen by iteratively searching for the cutoff that optimized the segment length predictions. The cutoff was defined as a frequency region below which the signal remains unchanged and above which the signal is removed. The optimal cutoff frequencies were determined by the 3dLift model by first running the marker data for each segment at cutoff frequencies from one to ten Hertz. At each set of frequencies for the i, j, and k directions, the average sum of the lengths between the markers defining the segment was calculated as the optimization criteria. At frequencies above the optimal cutoff frequency, it was assumed that the sum of the lengths between markers would increase as the amount of random noise in the data increased (see Figure 4.2). This assumption was valid if the noise was independent between individual markers, which means that position errors were statistically more likely to increase the sum of the lengths between markers than to decrease it. In addition, it was assumed that the sum of the lengths between markers would also increase below the optimal cutoff frequency as more and more of the actual motion data were removed. Expanding the example of Figure 4.2, the sum of the lengths may only increase by 0.1 mm for a segment when comparing the optimal cutoff frequencies to cutoff frequencies set at 10 Hz. However, a small error in position such as this occurring at a frequency of 10 Hz will be magnified to an acceleration error of approximately 0.4 m/s$^2$, which is significant for lifting motions (see Chapter Six).
Figure 4.2: Cutoff Frequency Optimization ($F_{ci} = 3$ Hz, $F_{cj} = 3$ Hz, $F_{ck} = 3$ Hz) (Taken from Trial Five, Asymmetric Leglift, Right Thigh Marker)
Table 4.2 shows the average cutoff frequencies chosen at the fifteen body segments for each lifting combination. Since the force platforms measure the combined forces of the entire body, their cutoff frequencies were conservatively set to the highest segment cutoff frequencies in each direction. Although optimal cutoff frequencies were set to the full range of two to ten Hertz when looking at individual trials, overall trends do not appear when the data are combined. The average cutoff frequencies in the i-direction were slightly lower (3.4 Hz) than in the j-direction (3.8 Hz) and k-direction (3.8 Hz). Similarly, when comparing average cutoff frequencies for different lifting combinations, the symmetric leglift (3.4 Hz) and symmetric backlift (3.4 Hz) were slightly lower than the asymmetric leglift (4.1 Hz) and asymmetric backlift (3.8 Hz). The overall segment frequencies ranged from 3.0 Hz at the left foot and upper torso to 4.0 Hz at the left hand and 4.6 Hz at the right hand. Although one might expect asymmetric lifts and motion at the hands to occur at higher frequencies, the differences were not substantial. Increased variability of the cutoff frequencies within individual trials was likely due not only to differences in motion, but also due to factors such as percentage of marker tracking.

In this chapter, methods of signal processing were discussed as a means of dealing with erroneous experimental data and missing data. Extrapolation and interpolation techniques were described to estimate missing experimental data that arose when markers became obscured. After filling in the missing points, noise residing in the experimental data was reduced using a Butterworth digital filter. Finally, the selection of an optimal cutoff frequency was outlined to remove as much of the noise as possible while leaving the lifting motion data intact. Once the signal processing was completed, the data were further analyzed to determine the segment orientations during the lifting trials. As developed in the next chapter, this involved determining the segment coordinate axes, the segment Cardan angles, and the segment joint centers and mass centers.
Table 4.2: Optimal Cutoff Frequencies (Hz) for Lifting Trials

<table>
<thead>
<tr>
<th>Segment</th>
<th>( F_c )</th>
<th>( F_{cl} )</th>
<th>( F_{ck} )</th>
<th>( F_c )</th>
<th>( F_{cl} )</th>
<th>( F_{ck} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot</td>
<td>4.0</td>
<td>4.8</td>
<td>3.0</td>
<td>4.7</td>
<td>3.7</td>
<td>3.3</td>
</tr>
<tr>
<td>Right Calf</td>
<td>2.0</td>
<td>2.0</td>
<td>4.5</td>
<td>2.0</td>
<td>4.7</td>
<td>4.7</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>3.8</td>
<td>2.0</td>
<td>6.0</td>
<td>3.3</td>
<td>3.0</td>
<td>7.7</td>
</tr>
<tr>
<td>Left Foot</td>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>2.3</td>
<td>2.3</td>
<td>4.3</td>
</tr>
<tr>
<td>Left Calf</td>
<td>2.0</td>
<td>6.0</td>
<td>4.5</td>
<td>2.0</td>
<td>2.7</td>
<td>6.3</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>3.3</td>
<td>5.3</td>
<td>3.0</td>
<td>4.7</td>
<td>2.0</td>
<td>5.0</td>
</tr>
<tr>
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<td>7.0</td>
<td>4.3</td>
<td>2.0</td>
<td>2.0</td>
<td>5.3</td>
<td>2.0</td>
</tr>
<tr>
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<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>4.7</td>
</tr>
<tr>
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<td>7.0</td>
<td>4.7</td>
<td>3.3</td>
<td>5.3</td>
</tr>
<tr>
<td>Right Forearm</td>
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<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>2.0</td>
<td>4.7</td>
</tr>
<tr>
<td>Right Hand</td>
<td>5.8</td>
<td>4.0</td>
<td>2.0</td>
<td>2.0</td>
<td>4.7</td>
<td>7.3</td>
</tr>
<tr>
<td>Left Upper Arm</td>
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<td>6.0</td>
<td>3.3</td>
<td>4.0</td>
<td>3.7</td>
<td>2.0</td>
</tr>
<tr>
<td>Left Forearm</td>
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<td>2.0</td>
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<td>6.7</td>
<td>2.0</td>
</tr>
<tr>
<td>Left Hand</td>
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<td>4.7</td>
<td>7.3</td>
<td>2.0</td>
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<tr>
<td>Head</td>
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<td>2.0</td>
<td>2.3</td>
<td>7.0</td>
</tr>
<tr>
<td>Force Platforms</td>
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<td>9.3</td>
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<td>10.0</td>
<td>10.0</td>
</tr>
</tbody>
</table>

Symmetric Backlift (n=4) Asymmetric Backlift (n=4)

<table>
<thead>
<tr>
<th>Segment</th>
<th>( F_c )</th>
<th>( F_{cl} )</th>
<th>( F_{ck} )</th>
<th>( F_c )</th>
<th>( F_{cl} )</th>
<th>( F_{ck} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot</td>
<td>4.3</td>
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<td>7.0</td>
<td>2.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Right Calf</td>
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<td>6.8</td>
<td>2.0</td>
<td>2.3</td>
<td>5.8</td>
</tr>
<tr>
<td>Right Thigh</td>
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<td>5.0</td>
<td>2.5</td>
<td>3.3</td>
<td>5.0</td>
</tr>
<tr>
<td>Left Foot</td>
<td>3.8</td>
<td>4.8</td>
<td>2.5</td>
<td>4.0</td>
<td>4.3</td>
<td>2.3</td>
</tr>
<tr>
<td>Left Calf</td>
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<td>3.3</td>
<td>2.0</td>
<td>6.5</td>
<td>3.8</td>
</tr>
<tr>
<td>Left Thigh</td>
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<td>4.8</td>
<td>2.0</td>
<td>2.0</td>
<td>4.3</td>
</tr>
<tr>
<td>Lower Torso</td>
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<td>2.0</td>
<td>5.5</td>
<td>3.3</td>
<td>2.0</td>
</tr>
<tr>
<td>Upper Torso</td>
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<td>4.0</td>
<td>2.3</td>
<td>4.0</td>
<td>2.0</td>
</tr>
<tr>
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</tr>
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<td>2.0</td>
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<td>3.8</td>
<td>5.3</td>
</tr>
<tr>
<td>Right Hand</td>
<td>5.3</td>
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<td>2.0</td>
<td>6.3</td>
<td>6.3</td>
<td>5.5</td>
</tr>
<tr>
<td>Left Upper Arm</td>
<td>2.5</td>
<td>3.0</td>
<td>3.3</td>
<td>5.3</td>
<td>7.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Left Forearm</td>
<td>2.0</td>
<td>2.0</td>
<td>4.0</td>
<td>4.5</td>
<td>8.0</td>
<td>4.0</td>
</tr>
<tr>
<td>Left Hand</td>
<td>3.0</td>
<td>2.0</td>
<td>2.0</td>
<td>6.5</td>
<td>6.8</td>
<td>2.0</td>
</tr>
<tr>
<td>Head</td>
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<td>7.0</td>
</tr>
<tr>
<td>Force Platforms</td>
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<td>9.5</td>
<td>10.0</td>
<td>10.0</td>
<td>9.8</td>
</tr>
</tbody>
</table>
CHAPTER FIVE: SEGMENT ORIENTATION

After signal processing, the marker data were now suitably accurate to analyze the three-dimensional orientation of the segments during the lifting trials. To begin, coordinate axes were determined for each of the fifteen body segments using the marker position data. This was a critical step, since coordinate axis information was utilized in developing the kinematics (Chapter Six) and equations of motion (Chapter Eight) of the body. Next, the angular positions of the segments in terms of Cardan angles were calculated using the segment coordinate axes. Following the changes in the Cardan angles over time allowed segment angular velocities and angular accelerations to be computed (Chapter Six). Segment joint centers and mass centers were found by combining subject anthropometry, marker position data, and segment coordinate axes. The segment joint center positions were needed for the equations of motion to estimate joint moments (Chapter Eight). In addition to also being needed to predict joint moments, the segment mass center positions were used to determine mass center accelerations (Chapter Six).

5.1 Segment Coordinate Axes

The 3dLift biomechanical model was composed of the fifteen segments shown in Figure 3.1, whose motion was tracked by the video markers listed in Table 3.3. A set of three orthogonal axes was created for each body segment, along with a global coordinate system defined by the video cameras. Figures 5.1 and 5.2 illustrate the segment coordinate axes and the markers associated with each segment. The alignment of the segment axes was consistent with extension-flexion in the sagittal plane, ad-abduction in the frontal plane, and twisting about the segment longitudinal axis (Cole et al., 1993). Therefore, when standing in the anatomical position, the i-axes of each segment were oriented predominantly in the anterior-posterior direction, the j-axes in the medial-lateral direction, and the k-axes in the vertical direction. Initially, a marker-based coordinate system was determined utilizing the marker combinations of each segment. The marker coordinate system consisted of two axes that were geometrically derived, and a third axis calculated to conform to a right-handed
Then, the marker coordinate system was transformed, if necessary, to the segment coordinate system to be consistent with the motion definitions.

Table 5.1 lists the steps taken to determine the marker coordinate axes, and Table 5.2 lists the conical angles from Table 3.2 used for transformation to the segment coordinate axes. A marker axis along the longest dimension of the segment was defined in step one by finding the relative position between two appropriate markers and normalizing this vector.

Normalization was carried out by dividing the vector by its magnitude, which was defined as the square root for the sum of the squared vector components. This initial marker axis was the $i_m$-axis for the feet, the $j_m$-axis for the lower and upper torso, and the $k_m$-axis for the remaining segments. A second marker axis was created by finding the normalized relative position between a different set of two segment markers and taking the cross product with the
first marker axis. The segments defined by two markers (upper arms, forearms, hands, and head) were treated differently, with their second marker axis defined by the relative shoulder marker positions. Shoulder marker positions were chosen as a reference since each of these segments could be traced back to the upper torso. A third marker axis was then calculated by taking the cross product of the first marker axis and the second marker axis.

Since the marker axes were often tilted with respect to the desired segment axes, a rotation of the axes by a conical angle was often required. For example, if a positive rotation about the $k_m$-axis was indicated in Table 5.2, then the transformation was carried out as follows:
Table 5.1: Marker Coordinate Axes

<table>
<thead>
<tr>
<th>Segment</th>
<th>Step One</th>
<th>Step Two</th>
<th>Step Three</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot</td>
<td>$i_{m1} = \frac{M_1 - M_3}{M_1 - M_3}$</td>
<td>$k_{m1} = \frac{i_{m1} \times (M_2 - M_1)}{i_{m1} \times (M_2 - M_1)}$</td>
<td>$j_{m1} = k_{m1} \times i_{m1}$</td>
</tr>
<tr>
<td>Right Calf</td>
<td>$k_{m2} = \frac{M_5 - M_3}{M_5 - M_3}$</td>
<td>$i_{m2} = \frac{(M_4 - M_4) \times k_{m2}}{(M_4 - M_4) \times k_{m2}}$</td>
<td>$j_{m2} = k_{m2} \times i_{m2}$</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>$k_{m3} = \frac{M_{11} - M_3}{M_{11} - M_3}$</td>
<td>$i_{m3} = \frac{(M_4 - M_4) \times k_{m3}}{(M_4 - M_4) \times k_{m3}}$</td>
<td>$j_{m3} = k_{m3} \times i_{m3}$</td>
</tr>
<tr>
<td>Left Foot</td>
<td>$i_{m4} = \frac{M_7 - M_8}{M_7 - M_8}$</td>
<td>$k_{m4} = \frac{i_{m4} \times (M_7 - M_6)}{i_{m4} \times (M_7 - M_6)}$</td>
<td>$j_{m4} = k_{m4} \times i_{m4}$</td>
</tr>
<tr>
<td>Left Calf</td>
<td>$k_{m5} = \frac{M_{10} - M_8}{M_{10} - M_8}$</td>
<td>$i_{m5} = \frac{(M_{10} - M_9) \times k_{m5}}{(M_{10} - M_9) \times k_{m5}}$</td>
<td>$j_{m5} = k_{m5} \times i_{m5}$</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>$k_{m6} = \frac{M_{12} - M_{10}}{M_{12} - M_{10}}$</td>
<td>$i_{m6} = \frac{(M_{10} - M_9) \times k_{m6}}{(M_{10} - M_9) \times k_{m6}}$</td>
<td>$j_{m6} = k_{m6} \times i_{m6}$</td>
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<tr>
<td>Lower Torso</td>
<td>$j_{m7} = \frac{M_{12} - M_{11}}{M_{12} - M_{11}}$</td>
<td>$i_{m7} = \frac{j_{m7} \times (M_{13} - M_{11})}{j_{m7} \times (M_{13} - M_{11})}$</td>
<td>$k_{m7} = i_{m7} \times j_{m7}$</td>
</tr>
<tr>
<td>Upper Torso</td>
<td>$j_{m8} = \frac{M_{19} - M_{15}}{M_{19} - M_{15}}$</td>
<td>$i_{m8} = \frac{j_{m8} \times (M_{14} - M_{13})}{j_{m8} \times (M_{14} - M_{13})}$</td>
<td>$k_{m8} = i_{m8} \times j_{m8}$</td>
</tr>
<tr>
<td>Right Upper Arm</td>
<td>$k_{m9} = \frac{M_{15} - M_{16}}{M_{15} - M_{16}}$</td>
<td>$i_{m9} = \frac{(M_{19} - M_{15}) \times k_{m9}}{(M_{19} - M_{15}) \times k_{m9}}$</td>
<td>$j_{m9} = k_{m9} \times i_{m9}$</td>
</tr>
<tr>
<td>Right Forearm</td>
<td>$k_{m10} = \frac{M_{16} - M_{17}}{M_{16} - M_{17}}$</td>
<td>$i_{m10} = \frac{(M_{19} - M_{15}) \times k_{m10}}{(M_{19} - M_{15}) \times k_{m10}}$</td>
<td>$j_{m10} = k_{m10} \times i_{m10}$</td>
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<tr>
<td>Right Hand</td>
<td>$k_{m11} = \frac{M_{17} - M_{18}}{M_{17} - M_{18}}$</td>
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<td>$k_{m15} = \frac{M_{23} - M_{14}}{M_{23} - M_{14}}$</td>
<td>$i_{m15} = \frac{(M_{19} - M_{15}) \times k_{m15}}{(M_{19} - M_{15}) \times k_{m15}}$</td>
<td>$j_{m15} = k_{m15} \times i_{m15}$</td>
</tr>
</tbody>
</table>
Table 5.2: Transformation from Marker Axes to Segment Axes

<table>
<thead>
<tr>
<th>Segment</th>
<th>Conical Angle</th>
<th>Rotation Axis</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot</td>
<td>$\psi_1$</td>
<td>$k_m1$</td>
</tr>
<tr>
<td>Right Calf</td>
<td>$-\psi_2$</td>
<td>$l_m2$</td>
</tr>
<tr>
<td>Right Thigh</td>
<td>$-\psi_3$</td>
<td>$l_m3$</td>
</tr>
<tr>
<td>Left Foot</td>
<td>$-\psi_4$</td>
<td>$k_m4$</td>
</tr>
<tr>
<td>Left Calf</td>
<td>$\psi_5$</td>
<td>$l_m5$</td>
</tr>
<tr>
<td>Left Thigh</td>
<td>$\psi_6$</td>
<td>$l_m6$</td>
</tr>
<tr>
<td>Lower Torso</td>
<td>$-\psi_7$</td>
<td>$l_m7$</td>
</tr>
<tr>
<td>Upper Torso</td>
<td>0</td>
<td>NA</td>
</tr>
<tr>
<td>Right Upper Arm</td>
<td>$-\psi_9$</td>
<td>$l_m9$</td>
</tr>
<tr>
<td>Right Forearm</td>
<td>$-\psi_{10}$</td>
<td>$l_m10$</td>
</tr>
<tr>
<td>Right Hand</td>
<td>$\psi_{11}$</td>
<td>$l_m11$</td>
</tr>
<tr>
<td>Left Upper Arm</td>
<td>$\psi_{12}$</td>
<td>$l_m12$</td>
</tr>
<tr>
<td>Left Forearm</td>
<td>$\psi_{13}$</td>
<td>$l_m13$</td>
</tr>
<tr>
<td>Left Hand</td>
<td>$-\psi_{14}$</td>
<td>$l_m14$</td>
</tr>
<tr>
<td>Head</td>
<td>$\psi_{15}$</td>
<td>$j_m15$</td>
</tr>
</tbody>
</table>

where $i_n$, $j_n$, and $k_n$ were segment coordinate axes; $\psi_n$ was the segment conical angle, and $i_mn$, $j_mn$, and $k_mn$ were marker coordinate axes. If the conical angle was negative, the effect was to switch the signs of the sine terms in the transformation matrix. As another example, the right forearm conical angle is depicted in Figure 5.2. The marker coordinate axes of the upper torso were already approximately aligned with the segment coordinate axes and thus no transform was required. Since the marker position data were obtained using video-based data, the segment coordinate axes were calculated in terms of global coordinates. The global coordinate axes were determined by the video camera calibration and were considered the fixed inertial reference frame in the 3dLift biomechancial model. This allows

\[
\begin{bmatrix}
i_n \\
j_n \\
k_n
\end{bmatrix} = \begin{bmatrix}
\cos \psi_n & \sin \psi_n & 0 \\
-\sin \psi_n & \cos \psi_n & 0 \\
0 & 0 & 1
\end{bmatrix} \begin{bmatrix}
i_mn \\
j_mn \\
k_mn
\end{bmatrix},
\] (5.1)
the segment coordinate axes to be rotating and translating reference frames while being referred back to the same fixed reference system.

5.2 Segment Cardan Angles

In human body modeling, simple mechanical equivalents have often been assumed to simulate the types of physical motion that occur at each of the anatomical joints. Most of the joints of interest in this research were classified as diarthroses or synovial joints, which were considered freely moving and capable of wide ranges of motion. For example, the ankles, knee, and elbows have previously been considered as hinge joints that restrict motion to one rotational degree of freedom about the medial-lateral axes. In addition, the wrists have been modeled as elliptical joints with two rotational degrees of freedom about the medial-lateral and anterior-posterior axes (Shrive and Frank, 1994). Furthermore, the hips and shoulders have been modeled as spherical joints with three rotational degrees of freedom with capability to rotate about each axis. In contrast, the intervertebral joints were referred to as amphiarthroses or cartilaginous joints and had limited ranges of motion if analyzed on an individual basis (Martini, 1992). However, when a section of multiple intervertebral joints were grouped together as in segments seven and eight (Figure 3.1), considerable mobility was achieved. Therefore, the combined intervertebral joints of the back and neck have been modeled as rotational gliding joints with three degrees of freedom and the ability to rotate about all axes.

In the 3dLift biomechanical model, segments defined by two markers (upper arms, forearms, hands, and head) were assumed to move with two rotational degrees of freedom. Since only two rotations could be resolved for two marker segments, these segments were restricted to flexion-extension and adduction-abduction. The remaining segments were analyzed with three rotational degrees of freedom to accommodate complex motions and as a check that the joints were behaving similarly to their mechanical equivalents. One method to represent the relative orientation of the segments is using Euler parameters to determine a unique axis about which a single rotation is performed (Amirouche, 1992). This was an attractive option in that it was linear in nature and sequence independent. However, additional transformations were required to interpret the results since the determined axes did
not necessarily have any physiological relevance. Another alternative was the use of helical angles to interpret movement from a reference position in terms of a rotation about and a translation along a directed line in space (Woltring, 1991). The 3dLift model used Cardan angles to describe the segment orientations in terms of three successive right-hand rotations because they could be easily related to physiological motion. Since Cardan angles involved multiplication of non-linear matrices and thus were order dependent, the rotation sequence was chosen to conform to the standard definitions of motion. As shown in Figure 5.3, the first joint rotation was flexion-extension about the j-axis, the second was adduction-abduction about the i-axis, and the third was axial rotation about the k-axis (Nigg, 1994).

The Cardan angles were determined by using the results for the segment coordinate axes outlined in Section 5.1. Each set of three segment coordinate axes was combined into a transformation or shifter matrix from the global coordinate system:

\[
\begin{bmatrix}
i_n \\
j_n \\
k_n
\end{bmatrix} =
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix} =
\begin{bmatrix}
i_{ni} & i_{nj} & i_{nk} \\
j_{ni} & j_{nj} & j_{nk} \\
k_{ni} & k_{nj} & k_{nk}
\end{bmatrix}
\begin{bmatrix}
i_{11} & S_{12} & S_{13} \\
S_{21} & S_{22} & S_{23} \\
S_{31} & S_{32} & S_{33}
\end{bmatrix}
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix},
\]

(5.2)

where \(i_n, j_n,\) and \(k_n\) were segment coordinate axes; \(i_{ni}\) to \(k_{nk}\) were global components of the segment coordinate axes, \(i, j,\) and \(k\) were global coordinate axes; and \(S_{11}\) to \(S_{33}\) were shifter matrix components. The two degree of freedom Cardan angle matrix was derived by multiplying an initial rotation about the global j-axis with a second rotation about an intermediate \(i'\)-axis:

\[
\begin{bmatrix}
i_n \\
j_n \\
k_n
\end{bmatrix} =
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix} =
\begin{bmatrix}
1 & 0 & 0 \\
0 & \cos \theta_n & \sin \theta_n \\
0 & -\sin \theta_n & \cos \theta_n
\end{bmatrix}
\begin{bmatrix}
\cos \phi_n & 0 & -\sin \phi_n \\
\sin \phi_n \sin \theta_n & \cos \theta_n & \cos \phi_n \sin \theta_n \\
\sin \phi_n \cos \theta_n & -\sin \theta_n & \cos \phi_n \cos \theta_n
\end{bmatrix}
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix},
\]

(5.3a)

\[
\begin{bmatrix}
i_n \\
j_n \\
k_n
\end{bmatrix} =
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix} =
\begin{bmatrix}
\cos \phi_n & 0 & -\sin \phi_n \\
\sin \phi_n \sin \theta_n & \cos \theta_n & \cos \phi_n \sin \theta_n \\
\sin \phi_n \cos \theta_n & -\sin \theta_n & \cos \phi_n \cos \theta_n
\end{bmatrix}
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix},
\]

(5.3b)

where \(\phi_n\) was the segment flexion-extension angle, and \(\theta_n\) was the segment adduction-abduction angle. For a three degree of freedom segment, the Cardan angle matrix included a third rotation about the segment k-axis:
Initial (Anatomical) Position

Flexion-Extension Angle

Adduction-Abduction Angle

Axial Rotation Angle

Figure 5.3: Three Degree of Freedom Segment Cardan Angles

\[
\begin{bmatrix}
  i_n \\
  j_n \\
  k_n
\end{bmatrix} =
\begin{bmatrix}
  \cos \phi_n \cos \beta_n + \sin \phi_n \sin \theta_n \sin \beta_n & \cos \theta_n \sin \beta_n \\
  -\cos \phi_n \sin \beta_n + \sin \phi_n \sin \theta_n \cos \beta_n & \cos \theta_n \cos \beta_n \\
  \sin \phi_n \cos \theta_n & -\sin \theta_n \\
  -\sin \phi_n \cos \beta_n + \cos \phi_n \sin \theta_n \sin \beta_n & \sin \phi_n \sin \beta_n + \cos \phi_n \sin \theta_n \cos \beta_n & \cos \theta_n \cos \beta_n
\end{bmatrix}
\begin{bmatrix}
  i \\
  j \\
  k
\end{bmatrix},
\]

(5.4)

where \( \beta_n \) was the segment axial rotation angle. Segment angular positions were calculated by comparing the experimental shifter matrices with the theoretical Cardan angle matrices and selected results appear in Appendix E. Using Equations (5.2) and (5.3), the flexion-extension angle and the adduction-abduction angle for two degree of freedom segments were simply calculated as shown:
\[
\phi_n = \tan^{-1}\left(\frac{\sin \phi_n}{\cos \phi_n}\right) = \tan^{-1}\left(\frac{-S_{13}}{S_{11}}\right), \quad (5.5a)
\]
\[
\theta_n = \tan^{-1}\left(\frac{\sin \theta_n}{\cos \theta_n}\right) = \tan^{-1}\left(\frac{-S_{32}}{S_{22}}\right). \quad (5.5b)
\]

Utilizing trigonometric identities on Equations (5.2) and (5.4), flexion-extension, adduction-abduction, and axial rotation angles were computed for three degree of freedom segments:

\[
\phi_n = \frac{1}{2} \tan^{-1}\left(\frac{k_p(S_{21} + S_{13})}{k_p(S_{23} - S_{11})}\right) + \frac{1}{2} \tan^{-1}\left(\frac{k_n(S_{21} - S_{13})}{k_n(S_{23} + S_{11})}\right). \quad (5.6a)
\]
\[
\beta_n = \frac{1}{2} \tan^{-1}\left(\frac{k_p(S_{21} + S_{13})}{k_p(S_{23} - S_{11})}\right) - \frac{1}{2} \tan^{-1}\left(\frac{k_n(S_{21} - S_{13})}{k_n(S_{23} + S_{11})}\right). \quad (5.6b)
\]
\[
k_p = \frac{1}{-S_{32} - 1}, \quad k_n = \frac{1}{-S_{32} + 1}, \quad (5.6c)
\]
\[
\theta_n = \tan^{-1}\left(\frac{\sin \theta_{n-1}}{\cos \theta_{n-1}}\right) = \tan^{-1}\left(\frac{-S_{32}}{S_{33} / \cos \phi_n}\right). \quad (5.6d)
\]

where \(k_p\) and \(k_n\) preserved the sign of the numerator and denominator for the ATAN2 function. In order to determine a unique angular solution and avoid gimbal lock or division by zero, the mathematical operator ATAN2 was used in both Equations (5.5) and (5.6).

### 5.3 Segment Joint Centers and Mass Centers

One of the main goals of the 3dLift model was the accurate determination of joint moments, and the location of the segment joint centers was critical for this analysis. Since most joints have been simplified as hinge or spherical joints, a common assumption has been that the joint centers occupy fixed positions relative to their adjacent segments. In the Hanavan (1964) geometric model of the human body, most joint centers were assumed to be located along the longitudinal axis at the intersection of articulating segments. For example, the knee joint was referenced to the calf, the elbow joint to the forearm, and the wrist joint to the hand. Using a subset of the Chandler et al. (1975) cadaver data, de Leva (1996b) provided estimates of joint center locations using anatomical landmarks as references. Joint centers were again assumed to be fixed relative to their adjacent segments, and they were
also positioned on the longitudinal axes of the segments. Although both sets of joint center locations are developed under similar assumptions, the Hanavan estimates appeared to be primarily based on convenient geometric terms. In contrast, the de Leva predictions seemed to have a stronger anatomical foundation and therefore were expanded upon and used in the 3dLift model.

Table 5.3 lists the parameters used to locate the positions of the segment joint centers and refers to Table 3.1, Table 3.3, and the segment coordinate axes of Section 5.1. Additional information on the sources used to develop the equations for locating the joint centers appears in Appendix C. The variable $R_M$ in the table represented the video marker radius, which was one-half inch for the lifting trials. Marker positions determined in terms of the global coordinate system served as reference points for the segment joint centers as listed in column two. The segment joint centers were located with respect to the marker positions through the adjustments listed in columns four to six along the segment axes listed in column three. Adjustments along the i-axes for the T10/T11 intervertebral joint and the neck joint were estimated using results from the computerized tomography study of Pearsall et al. (1996). With the exception of the T10/T11 joint and neck joint, the adjustments along the j-axes corresponded to a joint center located on the longitudinal axis of the segment. Again excluding the T10/T11 joint and neck joint, the adjustments along the k-axes were determined using the findings of de Leva (1996b). Since the marker positions were measured in global coordinates and the adjustments were made in segment coordinates, a transform was required using the segment axes:

$$I_n = M_{In} + [L_{in} L_{jn} L_{kn}] \begin{bmatrix} i_{in} & i_{jn} & i_{kn} \\ j_{in} & j_{jn} & j_{kn} \\ k_{in} & k_{jn} & k_{kn} \end{bmatrix} \begin{bmatrix} i \\ j \\ k \end{bmatrix},$$

(5.7)

where $I_n$ was the segment joint center for joint $n$, $M_{In}$ was the joint center reference marker, $L_{in}$, $L_{jn}$, and $L_{kn}$ were joint center adjustments; $i_{in}$ to $k_{kn}$ were joint center segment axis coordinates, and $i$, $j$, and $k$ were global coordinate axes.

The segment mass center locations were used to determine mass center accelerations, which in turn were utilized in the equations of motion. One alternative for estimating the mass center was to assume that it lies at the geometric center of the segment. This method
Table 5.3: Location of Segment Joint Centers

<table>
<thead>
<tr>
<th>Joint (Symbol)</th>
<th>Reference Marker</th>
<th>Segment Axes</th>
<th>Location i</th>
<th>Location j</th>
<th>Location k</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Ankle (J1)</td>
<td>M3</td>
<td>η₂</td>
<td>0</td>
<td>A28 + RM</td>
<td>-0.016(A7-A5)</td>
</tr>
<tr>
<td>Right Knee (J2)</td>
<td>M5</td>
<td>η₃</td>
<td>0</td>
<td>A33 + RM</td>
<td>0.074(A9-A7)</td>
</tr>
<tr>
<td>Right Hip (J3)</td>
<td>M11</td>
<td>η₃</td>
<td>0</td>
<td>A34 + RM</td>
<td>0.007(A9-A7)</td>
</tr>
<tr>
<td>Left Ankle (J4)</td>
<td>M8</td>
<td>η₅</td>
<td>0</td>
<td>-A28 - RM</td>
<td>-0.016(A7-A5)</td>
</tr>
<tr>
<td>Left Knee (J5)</td>
<td>M10</td>
<td>η₆</td>
<td>0</td>
<td>-A33 - RM</td>
<td>0.074(A9-A7)</td>
</tr>
<tr>
<td>Left Hip (J6)</td>
<td>M12</td>
<td>η₆</td>
<td>0</td>
<td>-A34 - RM</td>
<td>0.007(A9-A7)</td>
</tr>
<tr>
<td>T10/T11 Vertebrae (J7)</td>
<td>M13</td>
<td>η₇</td>
<td>-0.733A₁₆ + RM</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Right Shoulder (J8)</td>
<td>M13</td>
<td>η₉</td>
<td>0</td>
<td>A29 + RM</td>
<td>-0.104A₂₁</td>
</tr>
<tr>
<td>Right Elbow (J9)</td>
<td>M16</td>
<td>η₉</td>
<td>0</td>
<td>A₃₀ + RM</td>
<td>0.043A₂₁</td>
</tr>
<tr>
<td>Right Wrist (J10)</td>
<td>M17</td>
<td>η₁₀</td>
<td>0</td>
<td>A₃₅ + RM</td>
<td>0.006A₂₄</td>
</tr>
<tr>
<td>Left Shoulder (J11)</td>
<td>M19</td>
<td>η₁₂</td>
<td>0</td>
<td>-A₂₈ + RM</td>
<td>-0.104A₂₁</td>
</tr>
<tr>
<td>Left Elbow (J12)</td>
<td>M20</td>
<td>η₁₂</td>
<td>0</td>
<td>-A₃₀ + RM</td>
<td>0.043A₂₁</td>
</tr>
<tr>
<td>Left Wrist (J13)</td>
<td>M21</td>
<td>η₁₃</td>
<td>0</td>
<td>-A₃₅ + RM</td>
<td>0.006A₂₄</td>
</tr>
<tr>
<td>Neck (J14)</td>
<td>M14</td>
<td>η₈</td>
<td>-1.135A₃₂ + RM</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>

allowed for complicated geometries, but assumed that the segment had constant density distribution. Another commonly used technique was to estimate that the mass center lies a percentage of the segment length from one of the joint centers along the longitudinal axis. This percentage was taken from cadaver or non-invasive scanning studies and assumed symmetry about the longitudinal axis, while allowing for variable density distributions. Researchers have also developed regression equations to find segment mass centers using subject measurements as variables. These equations were also based on cadaver or scanning
studies and allowed for variable densities, asymmetric segment geometries, and different body types. In the 3dLift model, the mass centers as a ratio of segment length were used since the regression equations did not appear to show marked improvement in accuracy.

Table 5.4 outlines the variables used in locating the positions of the segment mass centers and refers to the anthropometric measurements in Table 3.1 and the markers in Table 3.3. Additional information on the sources used to derive to the equations for locating the segment mass centers appears in Appendix C. A new parameter, $\varphi$, was introduced to take into account the tilt of the foot from the ankle to the ground. The segment mass centers were located relative to the reference marker positions through adjustments along the axes for the segment of interest. With the exception of the lower and upper torso, the adjustments along the j-axes (i-axes for the feet) located the mass centers along the longitudinal axis of the segments. Further adjustments for the feet, calves, and thighs were derived using appropriate results from de Leva (1996a), Vaughan et al. (1992), Hinrichs (1990), and Clauser et al. (1969). The lower torso, upper torso, and head mass centers were estimated using the studies of Pearsall et al. (1996) and Zatsiorsky and Seluyanov (1983). Finally, the mass center locations for the upper arms, forearms, and hands were developed from the research of de Leva (1996a), Hinrichs (1990), and Clauser et al. (1969). Using the same steps as with the joint centers, the mass center adjustments made in segment coordinates were transformed into global coordinates:

$$
\mathbf{CM}_n = \mathbf{CM}_n + \begin{bmatrix}
L_{CMi} & L_{CMj} & L_{CMk}
\end{bmatrix}
\begin{bmatrix}
i_{ni} & i_{nj} & i_{nk} \\
j_{ni} & j_{nj} & j_{nk} \\
k_{ni} & k_{nj} & k_{nk}
\end{bmatrix}
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix}
$$

(5.8)

where $\mathbf{CM}_n$ was the mass center for segment $n$, $\mathbf{CM}_n$ was the mass center reference marker, $L_{CM}$ terms were mass center adjustments, $i_{ni}$ to $k_{nk}$ were segment axis coordinates, and $i$, $j$, and $k$ were global coordinate axes.

This chapter described the analysis methods undertaken by the 3dLift biomechanical model to convert video marker data into three-dimensional segment orientations. The first step in this process involved determining coordinate axes from marker positions for each segment in the model. Next, the segment coordinate axes were compared with the global
Table 5.4: Location of Segment Mass Centers

<table>
<thead>
<tr>
<th>Segment (Symbol)</th>
<th>Reference Marker</th>
<th>Location $i$</th>
<th>Location $j$</th>
<th>Location $k$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right Foot (CM₁)</td>
<td>$M_3$</td>
<td>0.443$A_3 / \cos \theta - A_{28}$</td>
<td>$A_{28} + R_M$</td>
<td>0</td>
</tr>
<tr>
<td>Right Calf (CM₂)</td>
<td>$M_3$</td>
<td>0</td>
<td>$A_{28} + R_M$</td>
<td>0.597$(A_7 - A_5)$</td>
</tr>
<tr>
<td>Right Thigh (CM₃)</td>
<td>$M_5$</td>
<td>0</td>
<td>0</td>
<td>0.624$(A_9 - A_7)$</td>
</tr>
<tr>
<td>Left Foot (CM₄)</td>
<td>$M_8$</td>
<td>0.443$A_3 / \cos \theta - A_{28}$</td>
<td>$-A_{28} - R_M$</td>
<td>0</td>
</tr>
<tr>
<td>Left Calf (CM₅)</td>
<td>$M_8$</td>
<td>0</td>
<td>$-A_{28} - R_M$</td>
<td>0.597$(A_7 - A_5)$</td>
</tr>
<tr>
<td>Left Thigh (CM₆)</td>
<td>$M_{10}$</td>
<td>0</td>
<td>$-A_{33} - R_M$</td>
<td>0.624$(A_9 - A_7)$</td>
</tr>
<tr>
<td>Lower Torso (CM₇)</td>
<td>$M_{13}$</td>
<td>$-0.543A_{16} - R_M$</td>
<td>0</td>
<td>$-0.497(A_{14} - A_9)$</td>
</tr>
<tr>
<td>Upper Torso (CM₈)</td>
<td>$M_{13}$</td>
<td>$-0.576A_{16} - R_M$</td>
<td>0</td>
<td>0.486$(A_{18} - A_{14})$</td>
</tr>
<tr>
<td>Right Upper Arm (CM₉)</td>
<td>$M_{15}$</td>
<td>0</td>
<td>$A_{29} + R_M$</td>
<td>$-0.514A_{21}$</td>
</tr>
<tr>
<td>Right Forearm (CM₁₀)</td>
<td>$M_{16}$</td>
<td>0</td>
<td>$A_{30} + R_M$</td>
<td>$-0.402A_{24}$</td>
</tr>
<tr>
<td>Right Hand (CM₁₁)</td>
<td>$M_{17}$</td>
<td>0</td>
<td>$A_{35} + R_M$</td>
<td>$-0.813A_{26}$</td>
</tr>
<tr>
<td>Left Upper Arm (CM₁₂)</td>
<td>$M_{19}$</td>
<td>0</td>
<td>$-A_{29} - R_M$</td>
<td>$-0.514A_{21}$</td>
</tr>
<tr>
<td>Left Forearm (CM₁₃)</td>
<td>$M_{20}$</td>
<td>0</td>
<td>$-A_{30} - R_M$</td>
<td>$-0.402A_{24}$</td>
</tr>
<tr>
<td>Left Hand (CM₁₄)</td>
<td>$M_{21}$</td>
<td>0</td>
<td>$-A_{35} - R_M$</td>
<td>$-0.813A_{26}$</td>
</tr>
<tr>
<td>Head (CM₁₅)</td>
<td>$M_{14}$</td>
<td>$-A_{31} - R_M$</td>
<td>0</td>
<td>0.613$(A_2 - A_{18})$</td>
</tr>
</tbody>
</table>

coordinate axes to find the angular positions of the segments in terms of Cardan angles. Finally, the segment joint centers and segment mass centers were estimated by adjusting the marker position data using anthropometry and the derived segment coordinate axes. In the next chapter, the changes in segment orientations over time were considered to find three-dimensional kinematic values. Specifically, segment angular velocities, segment angular accelerations, and segment mass center accelerations were calculated.
CHAPTER SIX: THREE-DIMENSIONAL KINEMATICS

After determining the segment orientations at each point in time, the three-dimensional kinematics of the human body during the lifting trials were analyzed. Segment angular velocities were found by calculating the changes in the segment angular positions with respect to time. These segment angular velocities were used later when summing moments about the segment mass centers in the equations of motion (Chapter Eight). Next, the segment angular accelerations were found by examining the changes in the segment angular velocities with respect to time. As with the angular velocities, the segment angular accelerations were utilized when summing moments in the equations of motion. Finally, the mass center accelerations were computed by taking the second derivative of the mass center positions with respect to time. Segment mass center accelerations were hereafter input when summing forces as part of the equations of motion. In addition to playing a role in finding joint forces and moments, kinematics also provided useful quantitative information about the movement of the body during the lifting motions.

6.1 Segment Angular Velocities - Theory

As an initial step in determining the angular velocities for each segment, the derivatives of the Cardan angles with respect to time were calculated. Since discrete data points were being used, the first derivatives of the Cardan angles were estimated using the finite difference method (Miller and Nelson, 1990):

\[ \dot{x}_1 = \frac{-3x_1 + 4x_2 - x_3}{2t}, \quad (6.1a) \]

\[ \dot{x}_i = \frac{x_{i+1} - x_{i-1}}{2t} \quad (i = 2, \ldots, n-1), \quad (6.1b) \]

\[ \dot{x}_n = \frac{3x_n - 4x_{n-1} + x_{n-2}}{2t}, \quad (6.1c) \]

where \( \dot{x}_i \) was the Cardan angular velocity at point \( i \), \( x_i \) is the Cardan angle at point \( i \), \( t \) is the time interval between data points, and \( n \) is the number of data points in the lifting trial. An alternative forward difference equation (6.1a) was used for the first data point and a
backward difference equation (6.1c) was used for the last data point. The angular velocity for a two or three degree of freedom body segment was expressed using the derivatives of the Cardan angles:

\[
\omega_n = \dot{\phi}_n j + \hat{\phi}_n i_n \quad \text{(two degrees of freedom), \quad (6.2a)}
\]

\[
\omega_n = \dot{\phi}_n j + \hat{\phi}_n i_n + \hat{\beta}_n k_n \quad \text{(three degrees of freedom), \quad (6.2b)}
\]

where \(\omega_n\) was the angular velocity of segment \(n\), \(\dot{\phi}_n\) was the flexion-extension angular velocity, \(\hat{\phi}_n\) was the adduction-abduction angular velocity, \(\hat{\beta}_n\) was the axial angular velocity, \(j\) was a global coordinate axis, \(i_n\) was an intermediate coordinate axis, and \(i_n\) and \(k_n\) were segment coordinate axes. Since the Cardan angles were defined as three successive rotations, these angles occurred about multiple reference frames (Andrews, 1995). In order to analyze multiple segments on the same basis, the angular velocity terms were transformed to the global coordinate system. Using the Cardan angle matrix of Equation (5.3), the angular velocities for two degree of freedom segments were found:

\[
\omega_n = \dot{\theta}_n \cos \phi_n i + \dot{\phi}_n j - \hat{\phi}_n \sin \phi_n k, \quad (6.3)
\]

where \(\phi_n\) was the segment flexion-extension angle, and \(\theta_n\) was the adduction-abduction angle. For three degree of freedom segments, the angular velocities were determined using the Cardan angle matrix of Equation (5.4):

\[
\omega_n = \left(\dot{\theta}_n \cos \phi_n + \hat{\beta}_n \sin \phi_n \cos \theta_n\right) i + \left(\dot{\phi}_n - \hat{\beta}_n \sin \theta_n\right) j + \left(- \dot{\theta}_n \sin \phi_n + \hat{\beta}_n \cos \phi_n \cos \theta_n\right) k, \quad (6.4)
\]

where \(\beta_n\) was the axial rotation angle.

### 6.2 Segment Angular Velocities - Results

Figure 6.1 compares the flexion-extension angular velocities and Figure 6.2 compares the axial rotation angular velocities for the thighs, lower torso, and upper torso. The Cardan angle derivatives are displayed due to their direct physiological interpretation, with appropriate trials grouped and averaged starting at the initiation of lifting. Referring to Figure 6.1, the flexion-extension angular velocities were greater at the thighs for the leglifts than the backlifts as is expected. The torso flexion-extension angular velocities were greater
Figure 6.1: Flexion-Extension Angular Velocities for Leglifts and Backlifts
(Leglifts - 7 trials, Backlifts - 8 trials)
Figure 6.2: Axial Rotation Angular Velocities for Leglifts and Backlifts
(Symmetric Leglifts – 4 Trials, Asymmetric Leglifts – 3 Trials,
Symmetric Backlifts – 4 Trials, Asymmetric Backlifts – 4 Trials)
during backlifts, but the difference was not as pronounced, indicating that back motion was required regardless of lifting style. Some variability existed between the right and left thighs, which may indicate a preferential leg or may have been manifested during the asymmetrical lifts. Asymmetry may also have been experimentally introduced due to obscuring of markers. An oscillation appeared in the upper torso flexion-extension angular velocity for backlifts, which may be explained by an initial liftoff phase followed by a secondary task-oriented phase. Examining Figure 6.2, there were clear differences in the axial rotation angular velocities between symmetrical and asymmetrical lifts as one might predict. The axial rotation angular velocities were slightly lower during the backlifts, and could be attributed to more efficiently coupled motion.

6.3 Segment Angular Accelerations - Theory

In order to determine the angular accelerations for each segment, the second derivatives of the Cardan angles with respect to time were required. Similar to the angular velocities, the second derivatives of the Cardan angles were calculated using the finite difference method (Miller and Nelson, 1990):

\[
\ddot{x}_i = \frac{2x_{i+1} - 5x_i + 4x_{i-1} - x_{i-2}}{t^2}, 
\]

\[
\ddot{x}_i = \frac{x_{i+1} - 2x_i + x_{i-1}}{t^2} \quad (i = 2, \ldots, n-1),
\]

\[
\ddot{x}_n = \frac{2x_n - 5x_{n-1} + 4x_{n-2} - x_{n-3}}{t^2},
\]

where \( \ddot{x}_i \) was the Cardan angular acceleration at point \( i \). A forward difference equation (6.5a) and a backward difference equation (6.5c) were again used for the first and last data points. The angular acceleration for two and three degree of freedom body segments included both changes in Cardan angular velocities and rotation of moving coordinate axes:

\[
\alpha_n = \dot{\phi}_n + \dot{\theta}_n + \dot{\beta}_n \quad \text{(two degrees of freedom)},
\]

\[
\alpha_n = \dot{\phi}_n + \dot{\theta}_n + \dot{\beta}_n + \dot{\gamma}_n \quad \text{(three degrees of freedom)},
\]

where \( \alpha_n \) was the angular acceleration of segment \( n \), \( \dot{\phi}_n \) was the flexion-extension angular acceleration, \( \dot{\theta}_n \) was the adduction-abduction angular acceleration, \( \dot{\beta}_n \) was the axial rotation
angular acceleration, \( \dot{a}_n \) was the intermediate coordinate axis derivative, and \( \dot{a}_n \) and \( \ddot{a}_n \) were segment coordinate axis derivatives. One method of deriving the angular acceleration equations was to solve for the rotating coordinate axis derivatives by taking the derivative of the Cardan angle matrix (Amirouche, 1992). An equivalent derivation consisted of taking the derivatives of Equations (6.3) and (6.4) with respect to time since these were already in terms of fixed global coordinates. Using the latter method with Equation (6.3), the angular accelerations for two degree of freedom segments were found as follows:

\[
\alpha_n = \left( \ddot{\theta}_n \cos \phi_n - \dot{\theta}_n \dot{\phi}_n \sin \phi_n \right) \hat{i} + \ddot{\phi}_n \hat{j} + \left( -\ddot{\theta}_n \sin \phi_n - \dot{\theta}_n \dot{\phi}_n \cos \phi_n \right) \hat{k}.
\]  

(6.7)

Furthermore, the angular accelerations for three degree of freedom segments were determined using Equation (6.4) as shown:

\[
\alpha_n = \left( \ddot{\theta}_n \cos \phi_n - \dot{\theta}_n \dot{\phi}_n \sin \phi_n + \beta_n \sin \phi_n \cos \theta_n + \beta_n \dot{\phi}_n \cos \phi_n \cos \theta_n - \beta_n \dot{\theta}_n \sin \phi_n \sin \theta_n \right) \hat{i} \\
+ \left( \ddot{\phi}_n - \ddot{\phi}_n \sin \theta_n - \beta_n \dot{\theta}_n \cos \theta_n \right) \hat{j} \\
+ \left( -\ddot{\theta}_n \sin \phi_n - \dot{\theta}_n \dot{\phi}_n \cos \phi_n + \beta_n \cos \phi_n \cos \theta_n - \beta_n \dot{\phi}_n \sin \phi_n \cos \theta_n - \beta_n \dot{\theta}_n \cos \phi_n \sin \theta_n \right) \hat{k}.
\]  

(6.8)

6.4 Segment Angular Accelerations - Results

Flexion-extension angular accelerations and axial rotation angular accelerations appear in Figures 6.3 and 6.4 for the thighs, lower torso, and upper torso. As was the case with the angular velocities, Cardan angle second derivatives are shown and trials were grouped according to lifting style and lifting motion. Flexion-extension was defined earlier as rotation about the global j-axis, and axial rotation was defined as rotation about the segment k-axis. Referring to Figure 6.3, the thigh flexion-extension angular accelerations were greater during the leglift trials, and the torso angular accelerations were greater during the backlift trials. The highest magnitude flexion-extension angular accelerations were seen in the lower and upper torso at the initiation and early lifting motions during the backlift trials. These greater magnitude angular accelerations indicate that changes in flexion-extension motion occurred more rapidly when lifting the object from the ground using a backlift. The increased oscillations in the left thigh flexion-extension angular accelerations are troubling and likely originated due to poorer marker tracking on the left-hand side of the body. As shown in Figure 6.4, axial rotation angular accelerations were greater during asymmetrical
Figure 6.3: Flexion-Extension Angular Accelerations for Leglifts and Backlifts
(Leglifts - 7 trials, Backlifts - 8 trials)
Figure 6.4: Axial Rotation Angular Accelerations for Leglifts and Backlifts
(Symmetric Leglifts – 4 Trials, Asymmetric Leglifts – 3 Trials,
Symmetric Backlifts – 4 Trials, Asymmetric Backlifts – 4 Trials)
lifts than during symmetrical lifts as expected. Asymmetrical backlifts resulted in higher axial angular accelerations than asymmetrical leg lifts, which indicated a jerkier (less smooth) lifting motion.

6.5 Segment Mass Center Accelerations - Theory

After determining the angular velocities and angular accelerations, the next step in the 3dLift biomechanical model was to find the linear accelerations at the segment mass centers. One method for determining mass center accelerations under the assumption of rigid bodies has been through the use of relative accelerations (Kane and Levinson, 1985):

\[
\mathbf{\ddot{a}}_{\text{cm}} = \mathbf{\ddot{a}}_{\text{cm}} + \mathbf{\ddot{\omega}}_{n} \times (\mathbf{CM}_n - \mathbf{J}_n) + \mathbf{\omega}_n \times (\mathbf{CM}_n - \mathbf{J}_n)
\]

(6.9)

where \(\mathbf{\ddot{a}}_{\text{cm}}\) was mass center acceleration, \(\mathbf{\ddot{a}}_{\text{cm}}\) was joint center acceleration, \(\mathbf{\ddot{\omega}}_{n}\) was segment angular acceleration, \(\mathbf{\omega}_n\) was segment angular velocity, \(\mathbf{CM}_n\) was mass center position, and \(\mathbf{J}_n\) was the joint center position. The first term represented the translational acceleration, the second term the tangential acceleration, and the third term the normal acceleration of the segment mass center. This method involved selecting a segment with a known or measured acceleration as the origin of the kinematic chain and relating the other segments through the joint center connections. These steps were not used in this research since it included numerous sources of error, although this method might reduce effects of skin displacement errors. A simpler, but theoretically equivalent, technique was used in the 3dLift biomechanical model and involved taking the second derivative of the mass center positions. Since the mass center positions determined in Section 5.3 were referenced to the fixed global coordinate system, the mass center accelerations were calculated using Equation (6.5). Mass center velocities, which would be required for a work-energy analysis, could have been found in a similar manner using Equation (6.1). In addition to its simplicity, this method lends itself to easier incorporation of moving joint centers if so desired, although it suffers from skin displacement errors.

Prior to the calculation of any velocities or accelerations, the raw marker position data were digitally filtered using the methods described in Chapter Four. The finite differences in Equations (6.1) and (6.5) also attenuated high frequency components in the data. Filtering
became especially critical when determining derivatives because the amplitude of the noise increased in proportion to the frequency squared for accelerations (Woltring, 1995). Therefore, if too high of a cutoff frequency was chosen, the data became increasingly noisy as the velocities and accelerations were calculated with the finite differences. In addition, Butterworth filtering was not perfect, and some noise was allowed to pass, especially low frequency noise and noise in the transitional region around the cutoff. To reduce some of the noise amplification caused by the finite differences, the data were again filtered after taking the derivatives and before using the results in further calculations. In this case, the highest cutoff frequency of the three segment axes was used since the acceleration effects were multi-directional. The problem of noise amplification is illustrated in Figure 6.5, which shows the raw position and raw through multiple filtered acceleration data for the head vertex marker in trial one. This demonstrates that if the positional data are processed without being filtered, the high frequency noise may dominate the calculated values and obscure the actual accelerations (Winter, 1990).

6.6 Segment Mass Center Accelerations - Results

Comparisons of mass center acceleration magnitudes for leglifts versus backlifts are shown in Figure 6.6 and for symmetrical versus asymmetrical lifts in Figure 6.7. Referring to Figure 6.6, the thigh mass center accelerations were only slightly greater for leglifts than backlifts. Higher peak mass center accelerations were seen in the upper torso during backlifts and, surprisingly, the lower torso during leglifts. As depicted in Figure 6.7, the torso mass center accelerations were slightly greater with some sharper peaks in asymmetrical leglifts. Upper torso mass center accelerations were greater during asymmetrical backlifts, and lower torso mass center accelerations were slightly greater during symmetrical backlifts. The highest mass center accelerations were seen in the upper torso during asymmetrical backlifts, which may indicate an increased difficulty for this lifting combination. The mass center accelerations seemed to be a function of where the segments were located in the kinematic chain relative to the stationary feet in additional to the type of lift. Additional kinematic results comparing the different lifting combinations using the 3dLift biomechanical model appear in Appendix F.
Figure 6.5: Raw vs. Filtered (Cutoff 2 Hz) Head Marker Accelerations
Figure 6.6: Mass Center Accelerations for Leglifts and Backlifts
(Leglifts - 7 trials, Backlifts - 8 trials)
Figure 6.7: Mass Center Accelerations for Symmetrical and Asymmetrical Lifts
(Symmetric Leglifts – 4 Trials, Asymmetric Leglifts – 3 Trials,
Symmetric Backlifts – 4 Trials, Asymmetric Backlifts – 4 Trials)
This chapter outlined the determination of kinematic values for comparing lifting motions and for later use in estimating joint forces and moments in the equations of motion. First, the segment angular velocities were derived from the time derivatives of the Cardan angles and results confirmed greater flexion-extension velocities in the thighs during leglifts and the torso during backlifts. In addition, greater axial rotation angular velocities were seen during asymmetrical lifts than during symmetrical lifts. Next, the segment angular accelerations were calculated from the second time derivatives of the Cardan angles, with greater flexion-extension accelerations in the thighs during leglifts and the torso during backlifts. Again, the asymmetrical lifts proved to have greater axial rotation angular accelerations than the symmetrical lifts. Finally, the segment mass center accelerations were found by taking the second time derivative of the segment mass center positions. The comparison between lifting motions was not as well defined in this case, but the highest mass center accelerations were seen in the upper torso during asymmetrical backlifts. Kinematics served as an input to both formulations of the 3dLift model, and in the next chapter, force platform analysis was discussed as an input to the lower body model.
CHAPTER SEVEN: FORCE PLATFORM MEASUREMENTS

Force platforms were used to measure the ground reactions occurring at the feet during lifting, which were then used as inputs to the lower body formulation in the 3dLift model. Researchers have developed methods for testing the accuracy of force platforms, and such information was valuable in indicating the amount of error in the ground reaction data. Hall et al. (1996) devised a series of calibration tests to experimentally determine a cross-sensitivity matrix to evaluate cross talk for individual force platforms. A point loader was used to apply vertical forces, and a pulley rig system with a latch plate was used to apply shear forces to a platform. Blanksby et al. (1995) calibrated a force platform in a vertical position using a hydraulic ram acting through a precision load cell. This allowed precise testing for a multitude of force magnitudes when the force platform was in an unconventional orientation. Bobbert and Schamhardt (1990) investigated the accuracy of center of pressure measurements using a stylus to apply vertical forces at known locations on the platform. It was found that the errors increased as the force application moved closer to the corners of the platform and that the increases in error were attributed to a lack of platform deformation at these loadings.

7.1 Force Platform Accuracy

Calibration of the force platforms allowed the output voltages to be converted to three ground reaction forces \( F_{GRi}, F_{GRj}, \) and \( F_{GRk} \) and three moments \( M_{GRi}, M_{GRj}, \) and \( M_{GRk} \). For this research, \( F_{GRi} \) corresponded to forces in the anterior-posterior direction, \( F_{GRj} \) corresponded to forces in the medial lateral direction, and \( F_{GRk} \) corresponded to forces in the vertical direction. The forces of interest in the 3dLift model were the ground reaction forces applied to the feet, whereas the platform measured forces according to the law of equal and opposite reactions. It was also necessary to account for the force platform coordinate systems being rotated ninety degrees about the vertical axis from the global coordinate system of the video cameras:

\[
F_{GRi} = F_{pj}, \quad (7.1a)
\]
\[ F_{GRj} = F_{Pi}, \quad (7.1b) \]
\[ M_{GRi} = M_{Fj}, \quad (7.1c) \]
\[ M_{GRj} = F_{Pi}, \quad (7.1d) \]

where \( F_{GRi} \) and \( F_{GRj} \) were ground reaction forces, \( F_{Pi} \) and \( F_{Pj} \) were platform forces, \( M_{GRi} \) and \( M_{GRj} \) were ground reaction moments, and \( M_{Pi} \) and \( M_{Pj} \) were platform moments. In addition, the force platform output voltages were divided by a conversion factor to adjust for amplification:

\[ CF = 10^{-3} V_{excite} G_{amp}, \quad (7.2) \]

where \( CF \) was the conversion factor, \( V_{excite} \) was the excitation voltage in volts, and \( G_{amp} \) was the amplifier gain. With the AMTI force platforms, the excitation voltages were set to 10 V and the amplifier gains to 4000 by setting DIP switches. Using Equations (7.1) and (7.2), the force platform output voltages were then multiplied by a calibration matrix to account for cross talk sensitivity in the strain gages:

\[
\begin{bmatrix}
F_{GRi} \\
F_{GRj} \\
F_{GRk} \\
M_{GRi} \\
M_{GRj} \\
M_{GRk}
\end{bmatrix} = \frac{1}{CF}
\begin{bmatrix}
C_{11} & C_{21} & C_{31} & \cdots & C_{51} & C_{61} \\
C_{12} & C_{22} & C_{32} & \cdots & C_{52} & C_{62} \\
C_{13} & C_{23} & C_{33} & \cdots & C_{53} & C_{63} \\
\vdots & \vdots & \vdots & \ddots & \vdots & \vdots \\
C_{61} & C_{62} & C_{63} & \cdots & C_{66}
\end{bmatrix}
\begin{bmatrix}
V_{PFi} \\
V_{PFj} \\
V_{PFk} \\
V_{PMi} \\
V_{PMj} \\
V_{PMk}
\end{bmatrix},
\quad (7.3)
\]

where \( C_{11} \) to \( C_{66} \) were platform calibration factors, \( V_{PF} \) was platform force voltage, and \( V_{PM} \) was platform moment voltage. The calibration factors were specific to each force platform and were provided by AMTI for this research (AMTI, 1991).

Force platform deformations were initially of concern, especially if bending of the platforms could cause detectable rotation in the direction of the ground reaction force vectors. Using mechanics of materials theory, the plate bending equation was used to estimate deflection under a 150 lb rectangular load placed at the platform center (Boresi et al., 1993; Jawad, 1994). The maximum displacement under these conditions was \( 1.4 \times 10^{-3} \) m, and it was therefore assumed that platform bending was unlikely to have a major effect during the lifting trials. The AMTI force platforms used in this research have been
constructed to deal with erroneous signals that result from platform deformation or platform support deformation (Carignan and Cook, 1985). Their design included the platform being supported by load cells that contained tubular cylindrical columns with strain gauges mounted upon them. By positioning the strain gauges on opposite sides of each column, it was claimed that most of the cross talk due to platform deformation was canceled out with a Wheatstone bridge. The remaining force platform errors due to bending and other forms of cross talk were compensated for in the calibration matrix of Equation 7.3. Platform bending might be a factor to consider when loading high magnitudes of weight or when severe impacts are involved.

In order to gain a sense of the accuracy of the force platforms, a series of simple calibration tests were run and analyzed. These tests involved stacking three fifty-pound weights in nine different combinations on a platform, with two trials collected for each configuration. The platform amplifier was set to a low-pass cutoff frequency of 10 Hz, data were sampled at 120 samples per second for five seconds, and the results were digitally Butterworth filtered at a 10 Hz cutoff. Platform positions were defined as A being the center, B being the (+i, +j) corner, C being the (+i, -j) corner, D being the (-i, -j) corner, and E being the (-i, +j) corner. The vertical force ($F_{GRI}$) should have been 666 N, while the anterior-posterior force ($F_{GRI}$), the medial-lateral force ($F_{GRJ}$), and the twisting moment ($M_{GRj}$) should have been zero for each trial. Figure 7.1 summarizes the force and moment results of the accuracy tests and shows the experimental mean and standard deviation of the data about the expected values. Looking at the overall results of the tests, it appears that low magnitude forces had an average error of 1.0 N, and forces in the range of body weight had an average error of 1.5 N. The anterior-posterior and medial-lateral forces had errors below 2.5 N, the vertical forces were within 4.7 N of the test weight, and the twisting moments remained below 1.1 Nm.

The center of pressure was the location for the resultant of the vertical ground reaction force as seen by the force platform (Section 7.3). For each weight configuration on the force platform, the center of pressure was different and should have been equal to the centroid of the calibration weights:
Figure 7.1: Accuracy of Force Platform Forces and Moments
(Letters signify placement of the three calibration weights as described in the text. For example, AAA stands for weight 1, position A; weight 2, position A; weight 3, position A)
where $C_{Wi}$ and $C_{Wj}$ were anterior-posterior and medial-lateral centroids, $W_1$ to $W_3$ were the weights, and $i_1$ to $j_3$ were the distances from the center of the platform surface to the center of the weight base. Calculating centers of pressure was not an exact accuracy test, because errors were introduced for any small differences in weight placements between trials. Errors could also be introduced if the calibration weights were not of a constant density or did not have a smooth surface contacting the force platform. Figure 7.2 shows the experimental and expected center of pressures for each accuracy trial, with the size of the crosses representing the standard deviation of the testing data. Average centers of pressure were 1.9 mm away from the expected value in the $i$-direction and 0.6 mm away in the $j$-direction, with standard deviations of 0.1 mm in both cases. The highest center of pressure errors were approximately 4.6 mm in the $i$-direction and 3.3 mm in the $j$-direction. These errors occurred when all three weights were stacked at the edge of the platform, indicating a decrease in accuracy as the center of pressure moves away from the platform center. It should be noted that the center of pressure errors are likely to become greater at smaller levels of applied force, which could be a factor in situations such as gait studies.

### 7.2 Ground Reaction Forces

As previously mentioned, ground reaction forces derived from force platform data were needed as kinetic inputs to the lower body formulation in the 3dLift biomechanical model. This research utilized two force platforms in order to avoid the problems associated with splitting up the ground reactions forces between two feet positioned on the same platform (Davis and Cavanagh, 1993). Prior to the first lifting trial and after the last one, readings were taken on the unloaded force platforms to determine a baseline output voltage. The average of the pre- and post-baseline voltages for the six channels of each platform was
Figure 7.2: Center of Pressure Accuracy Tests
subtracted from the experimental values to account for zeroing and voltage drift. At this point, Equation (7.3) was applied to both platforms to determine the ground reaction forces, starting at the synchronization pulse received from the video tracking system. In addition to accounting for the differences in the platform and global coordinate systems (Equation 7.1), markers placed on the platforms were digitized to determine finer adjustments. Using the methods of Section 5.1, a coordinate system for markers placed on the (+i, -j) edge, (-i, +j) edge, and the (-i, -j) edge of the right force platform was calculated:

\[
\begin{align*}
\hat{i}_p &= \frac{M_{p1} - M_{p3}}{|M_{p1} - M_{p3}|}, \\
\hat{j}_p &= \frac{M_{p2} - M_{p3}}{|M_{p2} - M_{p3}|}, \\
\hat{k}_p &= k,
\end{align*}
\]

where \( \hat{i}_p \) to \( \hat{k}_p \) were coordinate axes for the force platforms, \( M_{p1} \) to \( M_{p3} \) were platform markers, and \( k \) was the global vertical axis. Under the assumption that the platform and global vertical axes were equivalent, a single rotation about the vertical axis aligned the two systems:

\[
\gamma = \tan^{-1}\left( \frac{i_p}{j_p} \right),
\]

\[
\begin{bmatrix}
i \\
j \\
k
\end{bmatrix} =
\begin{bmatrix}
\cos \gamma & -\sin \gamma & 0 \\
\sin \gamma & \cos \gamma & 0 \\
0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
i_p \\
j_p \\
k_p
\end{bmatrix},
\]

where \( \gamma \) was the platform rotation, \( i_p \) and \( j_p \) were platform axis components, and \( i \) to \( k \) were global coordinate axes.

Figure 7.3 shows the vertical ground reactions and Figure 7.4 shows the medial-lateral ground reactions measured by the force platforms during the different lifting combinations. As was the case in Chapter Six, the graphs represent combined trials for symmetric and asymmetric leglifts and backlifts. Referring to Figure 7.3, the combined vertical ground reaction began approximately at the subject’s lifting weight for the leglift trials, but the backlift trials began below this level. This indicates that the subject was leaning on the
Figure 7.3: Force Platform Vertical Ground Reactions
Figure 7.4: Force Platform Medial-Lateral Ground Reactions
lifting object prior to the backlift, possibly using the object to push off from the floor and help initiate what was perceived as a more difficult task. The leglifts appeared to have a sharper rise to the peak vertical ground reactions, and this may be attributed to a more rapid acceleration of the lifting object. Combined vertical ground reactions showed an increase as the object was accelerated, a decrease as the object decelerated, and ending values at the subject plus lifting object weight. The right and left vertical ground reactions were similar for the symmetrical backlifts, but the right platform was slightly greater for much of the symmetrical leglifts. However, the right vertical ground reactions (the side lifted to) were much greater than the left side at the finish of the asymmetrical lifts.

In Figure 7.4, the medial-lateral ground reaction magnitudes were much less than the vertical ground reactions, but were important for determination of shear forces at the ankle joints. The medial-lateral ground reactions increased during the first half to two-thirds of the lift and indicated a greater need for frictional base support at the feet at these times. In general, the medial-lateral ground reaction forces then declined for the remainder of the lifting motion. The right and left medial-lateral ground reactions nearly cancelled out during the lifts, consistent with overall side-to-side balance or stability. However, the unbalance was slightly greater and favored the right side when the weight was held there at the completion of the asymmetrical lifts. Anterior-posterior ground reaction forces were lower in magnitude for these trials since no pushing or pulling was involved. The anterior-posterior ground reaction forces measured the need for frictional forces in this direction for front-to-back balance and stability. Anterior-posterior, medial-lateral, and vertical ground reaction forces were all used in the equations of motion to sum forces and moments at the feet.

7.3 Centers of Pressure

The centers of pressure for each foot were the locations of the resultant vertical ground reaction forces as measured by the force platforms. Accurate determinations of the centers of pressure were critical in defining the point of application of the ground reaction forces for the computation of joint moments. McCaw and DeVita (1995) found that center of pressure errors of 0.5 and 1.0 cm produced average changes in resultant joint moments between 7 and 14%. Center of pressure errors were complicated by the fact that data from two measurement
systems, force platforms and video cameras, were being combined. Using the platform forces and moments in Equation (7.3), the centers of pressure with respect to the geometric centers of the platforms were:

\[
\text{COP}_{Pi} = \frac{-M_{Pj} + F_{Pi}d_{Pk}}{F_{Pj}} - d_{Pi},
\]

(7.7a)

\[
\text{COP}_{Pj} = \frac{M_{Pj} + F_{Pj}d_{Pj}}{F_{Pj}} - d_{Pj},
\]

(7.7b)

where COP_{Pi} and COP_{Pj} were center of pressures \((\text{COP}_{Pk} = 0)\), \(M_{Pi}\) and \(M_{Pj}\) were platform moments, \(F_{Pi}\) to \(F_{Pj}\) were platform forces, and \(d_{Pi}\) to \(d_{Pj}\) were distances from the platform geometric origins to the combined strain gauge origins (AMTI, 1991). The free moments or the twisting moments of the feet about the vertical axis were then calculated using the platform center of pressures:

\[
M_{Fk} = M_{Pk} + F_{Pi}\text{COP}_{Pj} - F_{Pj}\text{COP}_{Pi},
\]

(7.8)

where \(M_{Fk}\) was the free moment \((M_{Fj} = M_{Fj} = 0)\). In order to integrate the force platforms with the video cameras, the center of pressures were translated to the global coordinate system using platform marker data. Using Equation (7.6) for rotation of the platform axes, the center of pressures with respect to the global coordinate system were:

\[
\text{COP}_{iP} = M_{P3i} + (\text{COP}_{P2i} + 0.5L_{Pi} - R_{M})\cos\gamma - (\text{COP}_{P2j} + 0.5L_{Pj} - R_{M})\sin\gamma,
\]

(7.9a)

\[
\text{COP}_{jP} = M_{P3j} + (\text{COP}_{P2i} + 0.5L_{Pi} - R_{M})\cos\gamma + (\text{COP}_{P2j} + 0.5L_{Pj} - R_{M})\sin\gamma,
\]

(7.9b)

\[
\text{COP}_{2i} = M_{P3i} + (\text{COP}_{P2i} + 0.5L_{Pi} - R_{M})\cos\gamma - (\text{COP}_{P2j} + 1.5L_{Pj} - R_{M} + L_{G})\sin\gamma,
\]

(7.9c)

\[
\text{COP}_{2j} = M_{P3j} + (\text{COP}_{P2i} + 0.5L_{Pi} - R_{M})\sin\gamma + (\text{COP}_{P2j} + 1.5L_{Pj} - R_{M} + L_{G})\sin\gamma,
\]

(7.9d)

where COP_{Pi} to COP_{Pj} were right and left center of pressures, \(L_{Pi}\) and \(L_{Pj}\) were platform dimensions, \(R_{M}\) was marker radius, \(\gamma\) was platform rotation, and \(L_{G}\) was the gap between platforms.

Figure 7.5 shows the anterior-posterior centers of pressure and Figure 7.6 shows the medial-lateral centers of pressure for the lifting trials. These values are with respect to the global coordinate system. As before, the graphs are averages over multiple trials of symmetric leg lifts (\(n = 4\)), asymmetric leg lifts (\(n = 3\)), symmetric back lifts (\(n = 4\)), and
Figure 7.5: Anterior-Posterior Centers of Pressure
Figure 7.6: Medial-Lateral Centers of Pressure
asymmetric backlifts (n = 4). The time axis starts at lift initiation, with lifting durations remaining consistent throughout the trials. Free moments remained low for the trials and would be more prominent if cutting or twisting foot motions were required. Referring to Figure 7.5, the anterior-posterior centers of pressure shifted forward as the object was lifted off the ground and shifted backwards as the object was moved toward the body. The starting centers of pressure were further back for the backlifts, indicating that the subject may have placed his feet further away from the object for these types of lifts. Left and right centers of pressure remained similar during symmetrical lifts, while the weight bearing right side was shifted posterior to the left side at the end of asymmetrical lifts. Looking at Figure 7.6, the medial-lateral centers of pressure remained relatively constant for the right and left sides during the course of the lifts. However, there appeared to be a slightly greater center of pressure shift on the left side toward the right during the last third of the asymmetrical lifts.

This chapter described the analysis of force platform data, which were used as a kinetic input to the lower body formulation in the 3dLift biomechanical model. Several basic accuracy tests were outlined that gave a sense of the errors associated with the platforms and served as a check of the platform calibrations. Platform forces had average errors of 1.5 N and centers of pressure had average errors of 2 mm for these accuracy tests. Ground reaction forces were determined and represented the contact forces acting between the feet and the force platforms during the lifting trials. Combined vertical ground reaction forces showed a similar pattern across lifting motions, but the right platform had a greater value than the left at the end of the asymmetrical lifts from the left to the right. The centers of pressure were calculated and were defined as the points of application of the vertical ground reaction forces. Anterior-posterior centers of pressure shifted forward early and backward late in the lifts, with the right side shifting posterior to the left side at the end of the asymmetrical lifts. Free moments were also computed as the twisting moments acting between the feet and the force platforms during the lifts. In the next chapter, the ground reaction forces will be input into the sum of forces and moments, while the centers of pressure and free moments will be used in the sum of moments.
CHAPTER EIGHT: EQUATIONS OF MOTION

The ultimate goal of this research was to accurately predict joint forces and joint moments that occur in the body during lifting motions. In order to accomplish this task, the topics discussed in the first seven chapters were combined: anthropometry, signal processing, segment orientations, kinematics, and kinetics. These factors were synthesized in equations of motion that were determined for each segment in the 3dLift biomechanical model.

Newton-Euler equations were utilized to develop these relationships, which involved a sum of forces and a sum of moments at each segment. The entire body was modeled, rather than just the lower back, since a motion that appears safe for the back alone may actually cause injury to surrounding joints. First, a lower body formulation was derived starting with measured inputs at the force platforms and ending at the T10/T11 intervertebral joint. Next, an upper body formulation was created starting with a known lifted weight at the hands and finishing again at the T10/T11 intervertebral joint. These two formulations provide multiple estimates at the T10/T11 intervertebral joint, which were later compared as a means of validating the model (Chapter Nine).

8.1 Biomechanical Models

Many biomechanical models that examine lifting focus primarily on the lower back, specifically the lumbosacral spine, because of the high incidence of injuries in that area of the body. Seireg and Arvikar (1989) predicted spinal muscle forces using minimization of muscle forces, joint forces, and joint moments as merit criterion to solve the overdetermined problem. Chaffin and Andersson (1991) developed both static and dynamic analyses to predict spinal loading, including specialized models designed to model common occupational tasks. Ladin et al. (1991) used muscle activity surfaces to predict the recruitment order of low back muscles resulting from holding weights in different orientations. Jager and Luttmann (1992) used a dynamic spatial model to calculate moments and forces at the L5/S1 intervertebral disc during simulated asymmetrical materials handling. Granata and Marras (1993) used electromyography data as input to a low-back model that simulated spinal loading during dynamic, asymmetric lifting. Kong et al. (1998) used a finite element model...
combined with optimization to predict loading of the lumbar spine anatomy during static sagittal plane lifting. Although the 3dLift model does not include specific spinal components at this time, it also analyzed surrounding joints in order to predict potentially high loadings at areas other than the back.

Depending on the motion to be analyzed, more generalized biomechanical models for the human body have been created using numerous methods with varying degrees of complexity. Garg and Chaffin (1975) used a three-dimensional, static model to predict the hand strength at different positions for a seated operator. To further consider muscular contributions, Pedotti (1977) developed a model that provided joint moments and muscle lengths using force platform and video marker data. Crowninshield and Brand (1981) used endurance as an optimization parameter to solve the overdetermined problem of predicting the individual muscle forces from joint moments. Lindbeck and Arborelius (1991) used a semidynamic analysis that included dynamic force platform data as an input to an otherwise static model to predict joint moments. Sitoh et al. (1993) introduced a lifting model that, in addition to dynamic analysis, included psychophysical factors to indicate which segments a subject was concentrating on using. An et al. (1995) explored both linear and nonlinear optimization methods to predict the distribution of muscle forces from joint moments. The 3dLift model includes three-dimensional, dynamic motion of the human body and focuses on finding accurate joint forces and moments. It has also been developed in a systematic fashion so that changing the marker set or segment divisions can be done with relative ease for future study of different types of motion. The use of validation also gives the 3dLift model increased confidence in the quantitative values for the joint forces and joint moments.

The three predominant methods used in biomechanics for finding equations of motion are based on Newton-Euler equations, Lagrange’s equations, and/or Kane’s equations. Six Newton-Euler equations are obtained for a three-dimensional segment analysis, with three of these derived by summing forces and three derived by summing moments (Nigg, 1994). While Newton-Euler equations are usually the easiest to derive and understand, this method may lead to more equations than are necessary and to an indeterminate system. Lagrange’s method is based on the principal of virtual work. It involves finding the segment kinetic and potential energies and forming three equations for a three degree of freedom segment.
(Amirouche, 1992). A smaller set for equations of motion are derived with Lagrange’s method than with Newton-Euler equations, but nonworking joint forces cannot be solved for (Chao, 1986). Models based on Kane’s equations are developed by adding the generalized active forces and the generalized inertia forces resulting in three equations for a three degree of freedom segment (Kane and Levinson, 1985). Kane’s method tends to be the most efficient in creating compact equations, but as with Lagrange’s equations, does not easily account for nonworking forces at the joints (Zajac and Winters, 1990). Since this research involved the solving of an inverse dynamics problem and nonworking joint forces were of interest, Newton-Euler equations were developed for the 3dLift model.

8.2 Lower Body Formulation

Using Newton-Euler equations, forces were balanced by linear inertial terms and moments by rotational inertial terms, resulting in two vector or six scalar relationships for each segment. For each set of six segment equations, three unknown joint forces and three unknown joint moments were determined. Joint forces and joint moments were resultant measures of all the muscle, ligament, tendon, and bone forces that act adjacent to or across the joint of interest. The lower body formulation began at the foot segments with ground reaction forces, free moments, and foot kinematics as known inputs:

\[ \mathbf{E}_{GR1} + m_1 \mathbf{g} = \mathbf{F}_{J1} \]
\[ \mathbf{M}_{FI} + (\mathbf{COP}_1 - \mathbf{CM}_1) \times \mathbf{E}_{GR1} - (\mathbf{I}_1 - \mathbf{CM}_1) \times \mathbf{F}_{J1} - \mathbf{M}_{J1} = [I_{IT}] \alpha_1 + \omega_1 \times [I_{IT}] \omega_1, \]
\[ \mathbf{E}_{GR2} + m_4 \mathbf{g} = \mathbf{F}_{J4} - \mathbf{M}_{J4}, \]
\[ \mathbf{M}_{F2} + (\mathbf{COP}_2 - \mathbf{CM}_4) \times \mathbf{E}_{GR2} - (\mathbf{I}_4 - \mathbf{CM}_4) \times \mathbf{F}_{J4} - \mathbf{M}_{J4} = [I_{IT}] \alpha_4 + \omega_4 \times [I_{IT}] \omega_4, \]

where \( \mathbf{E}_{GR1} \) and \( \mathbf{E}_{GR2} \) were ground reactions (Equations 7.3, 7.6), \( m_1 \) and \( m_4 \) were foot masses (Table 3.4), \( \mathbf{F}_{J1} \) and \( \mathbf{F}_{J4} \) were unknown ankle joint forces, \( \alpha_{G1} \) and \( \alpha_{G4} \) were foot mass center accelerations (Section 6.3), \( \mathbf{M}_{FI} \) and \( \mathbf{M}_{F4} \) were free moments (Equation 7.8), \( \mathbf{CM}_1 \) and \( \mathbf{CM}_4 \) were mass center positions (Equation 5.8), \( \mathbf{COP}_1 \) and \( \mathbf{COP}_4 \) were centers of pressure (Equation 7.9), \( \mathbf{I}_1 \) and \( \mathbf{I}_4 \) were ankle joint centers (Equation 5.7), \( \mathbf{M}_{J1} \) and \( \mathbf{M}_{J4} \) were unknown ankle joint moments, \( \alpha_1 \) and \( \alpha_4 \) were foot angular accelerations (Equation
and \( \omega_1 \) and \( \omega_4 \) were foot angular velocities (Equation 6.4). Gravity was assumed to act along the negative \( k \)-axis of the global coordinate system at a rate of 9.81 m/s\(^2\):

\[ g = -g_k. \]  

(8.2)

The segment moments of inertia and products of inertia were conveniently defined in terms of segment coordinate axes, and therefore had to be transformed to global coordinate axes:

\[
[I_{nT}] = \begin{bmatrix}
    i_{ni} & i_{nj} & i_{nk} \\
    j_{ni} & j_{nj} & j_{nk} \\
    k_{ni} & k_{nj} & k_{nk}
\end{bmatrix} \begin{bmatrix}
    I_{nii} & I_{nij} & I_{nik} \\
    I_{nji} & I_{njj} & I_{njk} \\
    I_{nki} & I_{nkj} & I_{nkk}
\end{bmatrix},
\]  

(8.3)

where \([I_{nT}]\) was the segment inertia matrix in global coordinates, \( I_{nii} \) to \( I_{nkk} \) were segment moments and products of inertia (Tables 3.5, 3.6), and \( i_{ni} \) to \( k_{nk} \) were components of the segment coordinate axes (Equation 5.2).

Equivalent steps were taken when developing the equations of motion for the remaining segments of the lower body formulation. Equations of motion for the calf segments used ankle joint forces, ankle joint moments, and calf kinematics as known values:

\[
F_{j1} + m_2 g - F_{j2} = m_2 \ddot{a}_2, \]  

(8.4a)

\[
( J_1 - CM_2 ) \times F_{j1} + M_{j1} - ( J_2 - CM_2 ) \times F_{j2} - M_{j2} = [I_{2T}] \alpha_2 + \omega_2 \times [I_{2T}] \omega_2, \]  

(8.4b)

\[
F_{j4} + m_5 g - F_{j5} = m_5 \ddot{a}_5, \]  

(8.4c)

\[
( J_4 - CM_5 ) \times F_{j4} + M_{j4} - ( J_5 - CM_5 ) \times F_{j5} - M_{j5} = [I_{5T}] \alpha_5 + \omega_3 \times [I_{5T}] \omega_3, \]  

(8.4d)

where \( F_{j2} \) and \( F_{j5} \) were unknown knee joint forces and \( M_{j2} \) and \( M_{j5} \) were unknown knee joint moments. Similar to the calf segments, the thigh segments included knee joint forces, knee joint moments, and thigh kinematics as known parameters:

\[
F_{j2} + m_3 g - F_{j3} = m_3 \ddot{a}_3, \]  

(8.5a)

\[
( J_2 - CM_3 ) \times F_{j2} + M_{j2} - ( J_3 - CM_3 ) \times F_{j3} - M_{j3} = [I_{3T}] \alpha_3 + \omega_3 \times [I_{3T}] \omega_3, \]  

(8.5b)

\[
F_{j5} + m_6 g - F_{j6} = m_6 \ddot{a}_6, \]  

(8.5c)

\[
( J_5 - CM_6 ) \times F_{j5} + M_{j5} - ( J_6 - CM_6 ) \times F_{j6} - M_{j6} = [I_{6T}] \alpha_6 + \omega_6 \times [I_{6T}] \omega_6, \]  

(8.5d)

where \( F_{j3} \) and \( F_{j6} \) were unknown hip joint forces and \( M_{j3} \) and \( M_{j6} \) were unknown hip joint moments. To complete the lower body model, the lower torso segment combined
known right and left hip joint forces, hip joint moments, and lower torso kinematics:

\[ E_{j3} + E_{j6} + m_7 g - E_{j7} = m_7 a_{G7}, \quad (8.6a) \]

\[ (I_3 - CM_3) \times E_{j3} + M_{j3} + (I_6 - CM_6) \times E_{j6} + M_{j6} - (I_7 - CM_7) \times E_{j7} - M_{j7} = [I_{\theta 7}] \alpha_7 + \omega_7 \times [I_{\theta 7}] \omega_7, \quad (8.6b) \]

where \( E_{j7} \) was the unknown T10/T11 intervertebral joint force and \( M_{j7} \) was the unknown T10/T11 intervertebral joint moment. The T10/T11 intervertebral joint forces and moments from the lower body formulation were compared to the predictions of these same values from the upper body formulation (Chapter Nine).

Figure 8.1 shows the compression and anterior-posterior shear joint forces for the knee, and Figure 8.2 shows these joint forces for the hip, both transformed to segment coordinate axes. Medial-lateral shear joint forces were much smaller in magnitude for the lifting motions studied and are not displayed. Knee compression forces (calf coordinate axes) were similar for symmetric leglifts and backlifts, with positive values indicating compressive forces acting on the calf segments. The asymmetric leglifts and backlifts resulted in greater knee compression forces during the latter stages of the lift for the side lifted toward (the right side), with this difference more pronounced in the leglifts. Hip compression forces (thigh axes) had similar characteristics to the knee, but were lower in magnitude since the hips did not have to support the weight of the thighs. Positive hip compression forces indicated compressive forces acting on the thigh segments. The anterior-posterior shear forces acting on the knee were much greater during leglifts, with negative values indicating anterior shear forces acting on the calf. Hip anterior-posterior shear forces were also greater during leglifts, with posterior shear forces acting on the thigh and approaching the compression force magnitudes early in the lift. The high shear forces acting on the knee and hip during the leglifts were of concern because the joints probably have less injury tolerance to shear forces than to compressive forces. The joint forces at the T10/T11 intervertebral joint predicted from the lower body formulation will be focused on in Chapter Nine.

Figure 8.3 shows the flexion-extension and adduction-abduction moments for the knee, and Figure 8.4 shows these joint moments for the hip, again transformed to segment axes. Axial rotation joint moments had lower magnitudes than the other joint moments for the
Figure 8.1: Knee Joint Forces
Figure 8.2: Hip Joint Forces
**Figure 8.3: Knee Joint Moments**
Figure 8.4: Hip Joint Moments
lifting trials and are not displayed. Negative flexion-extension moments at the knee joint indicate flexor moments, while negative flexion-extension moments at the hip joint indicate extensor moments. The knee flexion-extension joint moments acting counterclockwise on the calf segments were much greater during backlifts, which initially seemed counterintuitive. However, the knee flexion-extension moments were needed to stabilize the joint during backlifts, while the competing factors of stabilization and driven motion occurred during leglifts. This canceling of moments did not appear in the hips, where the required flexion-extension joint moments were similar for both leglifts and backlifts. Greater counterclockwise flexion-extension moments occurred on the right thigh segments during asymmetrical lifts, which was the side the weight was lifted toward. Positive adduction-abduction moments indicate abductor moments in the right knee and right hip, while positive adduction-abduction moments indicate adductor moments in the left knee and left hip. The adduction-abduction joint moments were greater in magnitude at the hips than at the knees, with right and left sides acting in an inverse manner during symmetrical lifts. T10/T11 intervertebral moments resulting from the lower body formulation will be displayed and discussed in Chapter Nine.

8.3 Upper Body Formulation

The upper body formulation began where the hands contacted the lifting object and progressively moved through the segments to the T10/T11 intervertebral joint. A mechanical delay of 0.042 seconds was first introduced to account for the time between the initiation of the lifting motion to force application by the hands. The load was then linearly increased from zero to the full weight of the lifting object over a transitional period of 0.2 seconds:

\[
\begin{align*}
P_L &= 0, \quad t < t_0, \\
P_L &= 0.5 \left( t - t_d - t_0 \right) / t_1, \quad (t_0 + t_d) \leq t \leq (t_0 + t_d + t_t), \\
P_L &= 0.5, \quad t > (t_0 + t_d + t_t),
\end{align*}
\]

where \(P_L\) was load percentage, \(t\) was time, \(t_0\) was lift initiation time, \(t_d\) was delay time, and \(t_t\) was transition time. The delay time was determined from the average time between the initiation of increased vertex vertical marker position and the initiation in increased vertical
ground reaction forces. Furthermore, the transition time was determined as the average time between the delay time and the point at which the combined vertical ground reactions equaled the combined subject and lifting object weights. This applied load was divided evenly between the right and left hands while taking into account the acceleration of these segments:

\[ F_{L11} = -P_L m_L a_{G11} + P_L m_L g, \]  
\[ F_{L14} = -P_L m_L a_{G14} + P_L m_L g, \]

where \( F_{L11} \) and \( F_{L14} \) were the right and left hand loads, \( a_{G11} \) and \( a_{G14} \) were the hand mass center accelerations, and \( m_L \) was the lifting object mass. Therefore, the equations of motion for the hand segments combined these derived relationships for the applied load with experimentally measured hand kinematics:

\[ F_{L11} + m_{11} g - F_{J10} = m_{11} a_{G11}, \]  
\[ (I_{10} - CM_{11}) \times F_{J10} - M_{J10} = [I_{11T}] a_{11} + \omega_{11} \times [I_{11T}] \omega_{11}, \]
\[ F_{L14} + m_{14} g - F_{J13} = m_{14} a_{G14}, \]
\[ -(I_{13} - CM_{14}) \times F_{J13} - M_{J13} = [I_{14T}] a_{14} + \omega_{14} \times [I_{14T}] \omega_{14}, \]

where \( F_{J10} \) and \( F_{J13} \) were unknown wrist joint forces and \( M_{J10} \) and \( M_{J14} \) were unknown wrist joint moments. It was assumed that the hands did not apply a moment to the lifted load and that only forces existed at the hand/load interface. It was also assumed that the point of application of the lifted load was at the mass center of the hands.

As was the case with the lower body formulation, the equations of motion for the remaining segments in the upper body formulation followed the same systematic steps. The forearm segment equations of motion contained known wrist joint forces, wrist joint moments, and forearm kinematics:

\[ F_{J10} + m_{10} g - F_{J9} = m_{10} a_{G10}, \]  
\[ (I_{10} - CM_{10}) \times F_{J10} + M_{J10} - (I_{9} - CM_{10}) \times F_{J9} - M_{J9} \]
\[ = [I_{10T}] a_{10} + \omega_{10} \times [I_{10T}] \omega_{10}, \]
\[ F_{J13} + m_{13} g - F_{J12} = m_{13} a_{G13}, \]
where $F_{j9}$ and $F_{j12}$ were unknown elbow joint forces and $M_{j9}$ and $M_{j12}$ were unknown elbow joint moments. Much like the forearm segments, the upper arm segments involved elbow joint forces, elbow joint moments, and upper arm kinematics as known inputs:

$$F_{j9} + m_9 g - F_{j8} = m_9 \alpha_{G9}, \quad (8.11a)$$

$$F_{j12} + m_{12} g - F_{j11} = m_{12} \alpha_{G12}, \quad (8.11b)$$

$$F_{j12} - (I_{j12} - CM_{j12}) \times F_{j12} + M_{j12} = \left[ I_{12T} \right] \alpha_{12} + \omega_{12} \times \left[ I_{12T} \right] \omega_{12}, \quad (8.11c)$$

where $F_{j8}$ and $F_{j11}$ were unknown shoulder joint forces and $M_{j8}$ and $M_{j11}$ were unknown shoulder joint moments. To predict the neck joint forces and moments, the measured kinematic values served as the known parameters for the head segment:

$$m_{15} g - F_{j14} = m_{15} \alpha_{G15}, \quad (8.12a)$$

$$-(I_{j14} - CM_{j14}) \times F_{j14} - M_{j14} = \left[ I_{15T} \right] \alpha_{15} + \omega_{15} \times \left[ I_{15T} \right] \omega_{15}, \quad (8.12b)$$

where $F_{j14}$ was the unknown neck joint force and $M_{j14}$ was the unknown neck joint moment. Finally, the upper torso segment combined known shoulder joint forces, shoulder joint moments, neck joint forces, neck joint moments, and upper torso kinematics:

$$F_{j8} + F_{j11} + F_{j14} + m_8 g - F_{j7} = m_8 \alpha_{G8}, \quad (8.13a)$$

$$-(I_{j7} - CM_{j7}) \times F_{j7} + M_{j7} = \left[ I_{8T} \right] \alpha_{8} + \omega_{8} \times \left[ I_{8T} \right] \omega_{8}, \quad (8.13b)$$

where $F_{j7}$ was the unknown T10/T11 intervertebral joint force and $M_{j7}$ was the unknown T10/T11 intervertebral joint moment. Upon completion of the upper body formulation, the T10/T11 intervertebral joint forces were analyzed in relation to the calculations from the lower body formulation (Chapter Nine).

Figure 8.5 shows the compression and anterior-posterior shear forces for the elbow (forearm axes), and Figure 8.6 shows compression and anterior-posterior shear forces for the
Figure 8.5: Elbow Joint Forces
Figure 8.6: Shoulder Joint Forces
shoulder (upper arm axes). Negative compression forces at the elbow joint indicated tensile forces acting on the forearm segments, while negative anterior-posterior shear forces indicated forces acting in the anterior direction with respect to the forearm segments. The elbow compression forces were high when the arms were vertical at lift initiation, then shifted to greater anterior-posterior forces as the forearms were horizontal at lift completion. Elbow medial-lateral shear forces were comparatively small for the lifting trials and are not shown in the figure. Negative compression forces at the shoulder joint indicated tensile forces acting on the upper arm segments, while negative anterior-posterior shear forces indicated forces acting in the anterior direction with respect to the forearm segments. Compression forces at the shoulder were greater than at the elbow since the joint had to additionally withstand the weight of the upper arms. The anterior-posterior shear forces at the shoulder were lower than at the elbow since the upper arms remained nearly vertical. Anterior-posterior forces were greater at the left shoulder than at the right shoulder at the completion of the asymmetric lifts. This was caused by additional stretching at the left shoulder as the left arm had a longer reach to the lifting object being held on the right side of the body. The T10/T11 intervertebral forces that were predicted during the lifting trials using the upper body formulation are discussed in Chapter Nine.

Figure 8.7 shows the flexion-extension and adduction-abduction elbow moments, and Figure 8.8 shows the flexion-extension and axial twist shoulder moments. Positive flexion-extension moments at the elbow and shoulder joints indicated flexor moments. Positive adduction-abduction moments at the right elbow joint indicated abductor moments, while positive adduction-abduction moments at the left joint indicated adductor moments. Positive axial twist moments in the right shoulder joint indicated external rotator moments, while positive axial twist moments in the left shoulder joint indicated internal rotator moments. The flexion-extension moments had a sharp rise at lift initiation followed by a slower increase for both the elbow and shoulder, with the shoulder slightly greater in magnitude. Adduction-abduction moments in the elbows displayed approximately equal and opposite values for the right and left sides. Similarly, the axial twist moments in the shoulders resulted in approximately equal and opposite predictions for the right and left sides. The axial twist moments in the elbow and the adduction-abduction moments in the shoulder were
Symmetric Leglift
Flexion-Extension
Adduction-Abduction

Symmetric Backlift
Flexion-Extension
Adduction-Abduction

Asymmetric Leglift
Flexion-Extension
Adduction-Abduction

Asymmetric Backlift
Flexion-Extension
Adduction-Abduction

Figure 8.7: Elbow Joint Moments
Figure 8.8: Shoulder Joint Moments
lower in magnitude and are not displayed. Since the elbow and shoulder joints played similar roles in the lifts studied, little difference was seen between leglifts and backlifts or symmetrical and asymmetrical lifts. Additional joint forces and joint moments calculated using the lower and upper body formulations appear in Appendix G. Joint moments for the T10/T11 intervertebral joint as estimated by the upper body formulation are reviewed in Chapter Nine.

In this chapter, equations of motion were developed for the 3dLift biomechanical model that resulted in predictions for joint forces and joint moments while lifting. The lower body formulation began with measured inputs at the force platforms and proceeded through the body to the T10/T11 intervertebral joint. For the knee and hip joints, anterior-posterior shear forces were greater during the leglifts, and compression forces were greater in the right side during the asymmetrical lifts. Flexion-extension moments were greater during backlifts for the knee joints, and flexion-extension moments were greater during asymmetrical lifts for the right knee and hip joints. The upper body formulation started with a known lifting object weight and systematically moved through the body to the T10/T11 intervertebral joint. Compression forces shifted to anterior-posterior shear forces during the lift for the elbow joint, and compression forces were predominant throughout the lift in the shoulder joint. Flexion-extension moments were greatest in magnitude for both the elbow and shoulder joints, with the shoulder joints slightly higher in value. In the next chapter, results for the T10/T11 intervertebral joint are focused upon as a means of validation for the 3dLift biomechanical model.
CHAPTER NINE: DISCUSSION OF MODEL PERFORMANCE

The goal of this research was to develop a methodology for studying lifting motions that have been linked to an increased risk of lower back injuries. To accomplish this, equations of motion have been developed for the 3dLift biomechanical model in order to determine the joint forces and joint moments. In this chapter, the forces and moments at the T10/T11 intervertebral joint that were predicted using lower and upper body formulations are compared to validate the 3dLift model. Agreement between the models will result in confidence that the force and moment predictions were reasonably accurate throughout the body. As a method of studying differences between the lower and upper body models, sensitivity analysis is introduced to measure the relative contributions to the moment calculations. The sensitivity analyses are further broken down into effective errors as a means of identifying specific parameters that were likely to introduce discrepancies in the results.

9.1 Model Validation

One test to validate the 3dLift biomechanical model was to compare the force predictions for the lower and upper body formulations at the T10/T11 intervertebral joint. Figure 9.1 shows the T10/T11 intervertebral compression forces and Figure 9.2 shows the T10/T11 intervertebral shear forces (lower torso axes). The joint forces show good agreement, with an average absolute difference of 10.4 N for anterior-posterior shear, 3.0 N for medial-lateral shear, and 3.5 N for compression. Comparing these values to the average absolute maximums, the percentage differences were 2.5% for anterior-posterior shear, 8.8% for medial-lateral shear, and 1.0% for compression. The T10/T11 intervertebral compression forces steadily increased during the course of the lift as the back became more upright, with lower initial compressions seen in the backlifts. Anterior-posterior shear forces increased during the first quarter of the lift as the back was more horizontal and the load was applied, then declined as compression increased. Differences appeared between the lower and upper body models at the initiation of the backlifts. A likely explanation for this discrepancy is discussed shortly. The medial-lateral shear forces had lower magnitudes than the other joint forces.
Figure 9.1: T10/T11 Intervertebral Compression Forces
Figure 9.2: T10/T11 Intervertebral Shear Forces
forces and had slightly greater values during asymmetrical lifts, but would be better tested during lateral bending. As mentioned previously, joint forces must be interpreted with care, since they include the resultant of muscle, ligament, tendon, disc compression, and bone contact forces.

A second validation test for the 3dLift biomechanical model was to compare the T10/T11 intervertebral joint moments from the lower and upper body formulations. Figure 9.3 shows the flexion-extension moments and Figure 9.4 shows the adduction-abduction and axial twist moments. Unfortunately, the joint moments did not compare as favorably as the forces, with average absolute differences of 7.0 Nm for adduction-abduction, 21.1 Nm for flexion-extension, and 8.5 Nm for axial twist. Comparing these values to the average absolute maximums, the percentage differences were 61.4% for adduction-abduction, 25.1% for flexion-extension, and 83.3% for axial twist. The T10/T11 intervertebral flexion-extension moments increased during the first quarter of the lift and then decreased, with higher peaks reached during backlifts. Although the flexion-extension moment predictions shared similar curve characteristics, the upper body model was consistently greater in magnitude than the lower body model. The adduction-abduction moments had much lower magnitudes than the flexion-extension moments and may be better validated using one-handed lifting motions. Somewhat surprisingly, the axial twist moments were also much lower than the flexion-extension moments and may be better tested with simple trunk rotations. Again, these results must be carefully interpreted in that they were a resultant of a wide variety of physiological forces that were applying a moment about the joint of interest.

Since the predicted T10/T11 joint forces resulted in a close match between models, there exists confidence that the forces calculated at intermediate joints were also accurate. A practical aspect to the validation of both models is that the lower body can be used to predict T10/T11 forces when force platform data can be taken, but hand forces are unknown. Conversely, the upper body model can be used to predict T10/T11 forces when applied loads at the hands can be measured, but taking force platform data is not feasible. The one area where the forces did not correspond was in the anterior-posterior shear forces during the first quarter of the backlifts. An explanation of this discrepancy was depicted in the vertical ground reactions of Figure 7.3, which indicated that the subject was leaning on the lifting
Figure 9.3: T10/T11 Intervertebral Flexion-Extension Moments
Figure 9.4: T10/T11 Intervertebral Joint Moments
object prior to the backlifts. The subject may have done this to push off the ground and help initiate the lift or as a means of balance during an awkward starting position for the backlift. This problem can be corrected by measuring forces under the lifting object or by specifically requesting in the trial protocol that no downward force be applied to the object. The differences between the models for T10/T11 joint moment predictions are more complex and will be studied using sensitivity analysis.

9.2 Sensitivity Analysis

Although the T10/T11 joint moment estimations had a similar form for each model and were therefore encouraging, there existed discrepancies in the predicted magnitudes. In order to analyze these differences, sensitivity analyses were performed to determine which factors had the greatest effect on the moment calculations. A common point at which the lifting trials averaged the highest overall moment magnitude was chosen to examine these moment contributions. The lower torso and upper torso segments were analyzed by finding the average magnitude of each term in the equations of motion (Equations 8.6, 8.13) during the lifting trials. Additional information was gained by considering the lower body and upper body as systems, thereby ignoring the equal and opposite effects of surrounding joint forces and moments. The system terms were then broken down into individual parameters that were expected to have varying degrees of accuracy. Combining the expected accuracy with the contribution sensitivity to the T10/T11 joint moments, an effective percentage of error was derived for each parameter. These steps were used to identify critical parameters with the highest effective errors for further examination when making fine adjustments to the lower and upper body models.

Referring to Figure 9.5, the lower torso segment was dominated by the hip joint moments, which were offset by moments due to the hip joint forces and T10/T11 joint forces. For the lower body system, moments due to ground reactions had the greatest contribution, which were offset by moments due to segment weights and T10/T11 joint force moments. Moments caused by the weight of the feet, free moments, linear inertia terms, and rotational inertial terms had significantly less effect on the results. Looking at Table 9.1, the expected accuracy of the anthropometric parameters were estimated by comparing values from several
Figure 9.5: Sensitivity Analysis for Lower Body Model
Table 9.1: Predicted Error Sources in Lower and Upper Body Models

**a) Lower Body System**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Expected Accuracy</th>
<th>Effective Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thigh Masses</td>
<td>15 N</td>
<td>9.1 %</td>
</tr>
<tr>
<td>Lower Torso Mass</td>
<td>15 N</td>
<td>3.6 %</td>
</tr>
<tr>
<td>Lower Torso Center of Mass</td>
<td>18 mm</td>
<td>3.5 %</td>
</tr>
<tr>
<td>Calf Masses</td>
<td>8 N</td>
<td>2.3 %</td>
</tr>
<tr>
<td>Platform Centers of Pressure</td>
<td>10 mm</td>
<td>2.2 %</td>
</tr>
<tr>
<td>Platform Ground Reactions</td>
<td>15 N</td>
<td>1.5 %</td>
</tr>
<tr>
<td>Thigh Centers of Mass</td>
<td>18 mm</td>
<td>1.3 %</td>
</tr>
<tr>
<td>Foot Masses</td>
<td>4 N</td>
<td>0.9 %</td>
</tr>
<tr>
<td>Calf Centers of Mass</td>
<td>17 mm</td>
<td>0.1 %</td>
</tr>
<tr>
<td>Foot Centers of Mass</td>
<td>5 mm</td>
<td>0.1 %</td>
</tr>
</tbody>
</table>

**b) Upper Body System**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Expected Accuracy</th>
<th>Effective Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forearm Masses</td>
<td>4 N</td>
<td>2.6 %</td>
</tr>
<tr>
<td>Upper Arm Masses</td>
<td>4 N</td>
<td>2.1 %</td>
</tr>
<tr>
<td>Head Mass</td>
<td>8 N</td>
<td>2.0 %</td>
</tr>
<tr>
<td>Hand Masses</td>
<td>2 N</td>
<td>1.8 %</td>
</tr>
<tr>
<td>Upper Torso Center of Mass</td>
<td>16 mm</td>
<td>1.6 %</td>
</tr>
<tr>
<td>Upper Torso Mass</td>
<td>15 N</td>
<td>0.8 %</td>
</tr>
<tr>
<td>Applied Loads at Hands</td>
<td>1 N</td>
<td>0.6 %</td>
</tr>
<tr>
<td>Load Contact Point at Hands</td>
<td>6 mm</td>
<td>0.4 %</td>
</tr>
<tr>
<td>Upper Arm Center of Mass</td>
<td>24 mm</td>
<td>0.2 %</td>
</tr>
<tr>
<td>Head Center of Mass</td>
<td>7 mm</td>
<td>0.2 %</td>
</tr>
<tr>
<td>Hand Center of Mass</td>
<td>3 mm</td>
<td>0.2 %</td>
</tr>
<tr>
<td>Forearm Center of Mass</td>
<td>16 mm</td>
<td>0.1 %</td>
</tr>
</tbody>
</table>

literature sources (see Chapter Three). Furthermore, expected accuracy of the force platform parameters was estimated by taking into account the calibration weight testing in Chapter Seven. The thigh masses and the lower torso were the most likely to introduce error in the T10/T11 joint moments and should be examined for modeling accuracy and adjusted if necessary. These segment masses had the greatest effect both because of their high magnitudes and their relatively low expected accuracy. Errors potentially introduced by the
mass center positions tended to be highest for the segment adjacent to the joint of interest, with the sensitivity declining as one moves more distal in the body. Although the ground reactions and centers of pressure play important roles in the magnitude of the moments, their relatively high accuracy lowers the effective error of these parameters. It should be noted that the choice of joint centers were arbitrary in terms of matching the two models, although they become critical when partitioning the joint forces into physiological components.

In Figure 9.6, the moments due to shoulder, T10/T11, and neck joint forces combined with shoulder and neck joint moments to form the T10/T11 intervertebral joint moment. Analyzed as a system, the moments due to applied loads at the hands were supplemented by moments due to the T10/T11 joint forces, weight of the upper arms, and weight of the forearms. Factors such as moments due to the weight of the hands and head, linear inertia terms, and angular inertia terms had much less effect on the calculations. Referring to Table 9.1, the expected accuracy of the parameters were found as before, with the load accuracy reflecting the precision of the weighing scale. The forearm and upper arm masses were the most likely to cause differences between the two studies and merit further study or adjustment as necessary. These segment masses had the greatest effect due to their combined magnitudes and moment arms from the T10/T11 intervertebral joint. The upper torso center of mass position had the greatest effective error for a parameter that was not a segment mass. In general, the upper body system was less sensitive to error than the lower body system for the lifting motions tested. Sensitivity analysis is not restricted to the T10/T11 joint moments and does not have to be calculated at the point where moments are at or near their maximum values. It could be applied to the remaining joints in the body to fine tune other parameters and might be of value to study static postures to further simplify the forces and moments involved.

This chapter compared the results for the T10/T11 joint forces and moments from the lower and upper body formulations as a means of validating the 3dLift biomechanical model. The joint forces showed close agreement between formulations, with errors ranging from 1.0% for compression forces to 8.8% for medial-lateral shear forces. Joint moments showed similar patterns between models, but had a separation in magnitudes that corresponded to higher error percentages. To examine the differences in joint moments between
Figure 9.6: Sensitivity Analysis for Upper Body Model
formulations, sensitivity and effective error analyses were also introduced. Moments due to the ground reactions had the greatest effect on the lower body system, and moments due to the applied load had the greatest effect on the upper body system. The estimation of thigh masses and lower torso mass were a likely source of error for the lower body system, and the estimation of forearm and upper arm masses were a likely source of error for the upper body system.
CHAPTER TEN: CONCLUSIONS

Because of the prevalence of lower back injuries and the link to lifting activities as a common cause, this research was undertaken to study these types of motions. A subject performed a set of sixteen lifting trials while markers were tracked with video cameras, and ground reactions were measured by force platforms. Anthropometric measurements taken on the lifting subject individualized the 3dLift biomechanical model and provided parameters used throughout the research. Through a series of signal processing steps, missing marker data were estimated and errors in the data were attenuated. From the marker position data, the orientations of the body segments during the course of the lifting motion were determined. Next, three-dimensional kinematic quantities were calculated by examining how the segment orientations changed with respect to time. Finally, all these components were combined into equations of motion that made up a lower body model and an upper body model to estimate joint forces and moments. The T10/T11 intervertebral joint force and moment predictions from these two formulations were then compared to validate the 3dLift biomechanical model. While there is much work yet to be done in the area of this research, the 3dLift model takes the first steps by developing a systematic methodology for studying lifting motions. In this chapter, several of the research contributions of the 3dLift model are outlined. In addition, future areas of expansion for the model such as additional lifting trials and modeling of forces in individual anatomical structures are proposed.

10.1 Research Contributions

The 3dLift biomechanical model provides theoretical and experimental contributions in the areas of engineering mechanics and more specifically, multibody dynamics. Many lifting models are two-dimensional and static in nature due to the simplifications that such assumptions allow. The 3dLift model is three-dimensional, which allows more complex motions to be analyzed, such as lifts that involve both flexion-extension and axial twisting of the torso. Although kinematic results are valuable in themselves for analyzing motion, the dynamic aspects of the model did not have a significant influence on the kinetic terms for the lifting trials. However, the acceleration terms neglected in static analyses would likely
become more predominant as the weight lifted is increased and approaches levels where injuries begin to occur. The 3dLift model was also designed to be systematic in its structure, which allows changes to the model to be made with relative ease. Experimentally, the model introduces some basic steps for mathematically dealing with missing video marker data. Although it stills needs some refinement, the optimization of obscured marker positions using surrounding visible markers seems to be a particularly promising application.

The 3dLift model provides contributions in the areas of biomedical engineering and biomechanics through its experimental aspects and as a tool for injury prevention. Injuries to the back are common and costly, and the model provides further insight into the forces and moments that occur at the joints during lifting. The 3dLift model can be used to analyze a wide variety of lifting motions from railroad workers lifting manual switches to children manually lifting buckets on a farm. As more trials are analyzed and changes in kinematics with weight lifted become known, the model could also simulate lifts that have too high of a risk of injury for subjects to attempt. The 3dLift model includes both a lower body formulation and an upper body formulation, which gives additional experimental flexibility depending on the lift to be studied. For example, if the use of force platforms is not feasible, then the upper body formulation can still be used if loads at the hands are known. The two formulations also serve as a means of validating the 3dLift model, which shows close agreement for the T10/T11 joint forces and similar patterns for the T10/T11 joint moments. Different models can give very different results, but the validation methods of this research provide an experimental test of the quantitative reliability for the joint forces and moments.

10.2 Future Recommendations

To further evolve the 3dLift biomechanical model, a larger sample size of lifting trials need to performed and analyzed. Although not developed in this document, the results proved repeatable between trials involving the same lifting combinations for the subject tested. What was not studied in this research was the variability between subjects as the factors of individual anthropometry and differing lifting techniques come into play. In addition, more lifting situations need to be analyzed, such as one-handed lifting, lateral bending, and isolated axial rotations. Other factors of interest include different lifting object
starting and ending heights, different magnitudes of weight to be lifted, and variations in lifting velocity. The model could also be used in certain industrial settings if the lifting motion of concern is not easily repeatable in a laboratory setting. In such cases, model validation especially comes into play if either force platform measurements are not feasible or the force at the hands is not easily determined. As more lifting trials are documented, the accuracy of the 3dLift model should increase as sources of error are exposed and corrected when possible through model refinement.

Another area where the 3dLift biomechanical model can be improved is in terms of the experimental setup for capturing the lifting trials. The force platforms appeared to be acceptably accurate, with the exception of when forces were applied outside their surfaces, such as leaning on the lifting object. However, the video capture system introduced some problems of genuine concern in the form of obscured marker position data. When data were missing, the methods of Chapter Four were implemented to estimate the marker positions, which creates accuracy problems in addition to those listed in Table 9.1. Because of significant obscuring of a number of markers, they had to be dropped from the data, which reduced several segments to two degree of freedom analyses. Four video cameras were utilized for this research and placement was restricted on the left side of the body due to the laboratory configuration. The addition of a fifth and sixth camera along with a setup reconfiguration would allow three cameras to be focused on each side of the body and might significantly improve marker tracking. However, the original marker set may simply be too numerous for passive marker separation in the whole body volume and tracking the reduced marker set might be the best solution.

The one area where the 3dLift model relied heavily on previous research was in the determination of subject anthropometric parameters. While some parameters, such as segment masses, appeared to be in relative agreement in the literature, others, such as moments of inertia, were not. When examining Table 9.1, it was shown that joint moment calculations were sensitive to the position of the segment mass centers. In addition, if the model is to be extended to finding forces in individual anatomical structures (see below), then the position of the segment joint centers will become critical. For this research, anthropometric relationships were chosen to relate to the subject measurements and to show
some agreement among literature values. As more subject anthropometric data are collected, the simple equations used in the model may prove inadequate and more complex regression equations may be required. It may also be found that no set of equations provides a satisfactory match for all subjects, and fine adjustments of the parameters using model results might be in order (Section 9.2). In addition, the location of segment joint centers might be best found experimentally by relating linear velocities of markers on a segment and the angular velocities of that segment.

In another future development, the 3dLift model needs to incorporate the L5/S1 intervertebral disk, which has been documented as the disk most vulnerable to injury. The T10/T11 joint was studied in this research because of its positioning with respect to the reduced marker set and to split the human body model through the middle torso. Since the L5/S1 disk was not conveniently located with respect to the reduced marker set, greater errors would likely have occurred locating this joint than with other joints of the model. One of the primary goals of this research was initial validation of the model, and therefore using the L5/S1 disk as a joint center was not feasible at this point in the model development. In order to locate the L5/S1 joint center with the reduced marker set, new anthropometric equations need to be added to locate it with respect to the hip or substernale markers. The lower torso kinematics would also have to be partitioned, since the segment angular orientations were measured with divisions at the hips and at the substernale. If a revised marker set or camera configuration would allow the superior iliac markers to be properly tracked, then the difficulties with analyzing the L5/S1 joint center would be reduced. The L5/S1 joint forces and moments are expected to approach the combined joint forces and moments for the hips, which would likely be the highest values found in the body.

Finally, the 3dLift biomechanical model could be expanded in its scope to examine more specific factors regarding potential for injury. While joint forces and moments give a resultant value for loading at a joint, the individual forces on muscles, ligaments, discs, etc. would be a more reliable measure. This is not a simple task, as it requires a significant amount of additional anthropometric data and an assumed optimization method for the distribution of forces. This research focused on finding accurate joint force and moment predictions, which are naturally critical if the goal is to proceed and find accurate forces on
individual structures. In addition, criteria need to be developed to indicate at what force or moment magnitude levels that the subject at is risk of suffering an injury. For example, it is likely that injuries would occur at lower thresholds for joint shear forces than for joint compression forces. Such information may come from cadaveric studies, with injuries being reproduced experimentally. Another method of determining injury thresholds may come from the 3dLift model itself by simulating activities that statistics have shown cause injuries in the workplace.
APPENDIX A: INFORMED CONSENT

Appendix A contains the Information for Review of Research Involving Human Subjects form and the Informed Consent to Participate in Research form. These forms were approved by the Iowa State University Human Subjects Review Committee prior to the lifting trials being performed. The Informed Consent form was reviewed and signed by the lifting subject after any questions about the research were addressed and prior to the experiments.
Information for Review of Research Involving Human Subjects
Iowa State University
(Please type and use the attached instructions for completing this form)

1. Title of Project: A Three-Dimensional, Dynamic Model of the Human Body During Lifting

2. I agree to provide the proper surveillance of this project to insure that the rights and welfare of the human subjects are protected. I will report any adverse reactions to the committee. Additions to or changes in research procedures after the project has been approved will be submitted to the committee for review. I agree to request renewal of approval for any project continuing more than one year.

3. Signatures of other investigators

4. Principal investigator(s) (check all that apply)
   - Faculty
   - Staff
   - Graduate student
   - Undergraduate student

5. Project (check all that apply)
   - Research
   - Thesis or dissertation
   - Class project
   - Independent Study (490, 590, Honors project)

6. Number of subjects (complete all that apply)
   - # adults, non-students
   - # ISU students
   - # minors under 14
   - # minors 14 - 17

7. Brief description of proposed research involving human subjects: (See instructions, item 7. Use an additional page if needed.)

   See attached Informed Consent to Participate in Research.

8. Informed Consent:
   - Signed informed consent will be obtained. (Attach a copy of your form.)
   - Modified informed consent will be obtained. (See instructions, item 8.)
   - Not applicable to this project.
9. **Confidentiality of Data:** Describe below the methods you will use to ensure the confidentiality of data obtained. (See instructions, item 9.)

The principal investigators will assign a subject number to identify data files collected during this research, and the identity of the subject will be known only to the principal investigators. The subject's identity will not be associated with any of the findings presented as a result of this research project. Any video tapes obtained of the subject during this research will be erased after they have been analyzed unless permission has been granted by the subject to archive the video tapes. The video tapes will be used for research purposes only.

10. What risks or discomfort will be part of the study? Will subjects in the research be placed at risk or incur discomfort? Describe any risks to the subjects and precautions that will be taken to minimize them. (The concept of risk goes beyond physical risk and includes risks to subjects' dignity and self-respect as well as psychological or emotional risk. See instructions, item 10.)

   See attached Informed Consent to Participate in Research.

11. **CHECK ALL** of the following that apply to your research:

   - [ ] A. Medical clearance necessary before subjects can participate
   - [ ] B. Administration of substances (foods, drugs, etc.) to subjects
   - [X] C. Physical exercise or conditioning for subjects
   - [ ] D. Samples (blood, tissue, etc.) from subjects
   - [ ] E. Administration of infectious agents or recombinant DNA
   - [ ] F. Deception of subjects
   - [ ] G. Subjects under 14 years of age and/or □ Subjects 14 - 17 years of age
   - [ ] H. Subjects in institutions (nursing homes, prisons, etc.)
   - [ ] I. Research must be approved by another institution or agency (Attach letters of approval)

If you checked any of the items in 11, please complete the following in the space below (include any attachments):

**Items A–E** Describe the procedures and note the proposed safety precautions.

See attached Informed Consent to Participate in Research

**Items D–E** The principal investigator should send a copy of this form to Environmental Health and Safety, 118 Agronomy Lab for review.

**Item F** Describe how subjects will be deceived; justify the deception; indicate the debriefing procedure, including the timing and information to be presented to subjects.

**Item G** For subjects under the age of 14, indicate how informed consent will be obtained from parents or legally authorized representatives as well as from subjects.

**Items H–I** Specify the agency or institution that must approve the project. If subjects in any outside agency or institution are involved, approval must be obtained prior to beginning the research, and the letter of approval should be filed.
Checklist for Attachments and Time Schedule

The following are attached (please check):

12. □ Letter or written statement to subjects indicating clearly:
   a) the purpose of the research
   b) the use of any identifier codes (names, #s), how they will be used, and when they will be removed (see item 17)
   c) an estimate of time needed for participation in the research
   d) if applicable, the location of the research activity
   e) how you will ensure confidentiality
   f) in a longitudinal study, when and how you will contact subjects later
   g) that participation is voluntary; nonparticipation will not affect evaluations of the subject

13. ☒ Signed consent form (if applicable)

14. □ Letter of approval for research from cooperating organizations or institutions (if applicable)

15. □ Data-gathering instruments

16. Anticipated dates for contact with subjects:

<table>
<thead>
<tr>
<th>First contact</th>
<th>Last contact</th>
</tr>
</thead>
<tbody>
<tr>
<td>12/1/98</td>
<td>2/15/99</td>
</tr>
</tbody>
</table>

Month/Day/Year          Month/Day/Year

17. If applicable: anticipated date that identifiers will be removed from completed survey instruments and/or audio or visual tapes will be erased:

5/15/99
Month/Day/Year

18. Signature of Departmental Executive Officer: Date

[Signature]
1/16/98

Department or Administrative Unit
Aerospace Engineering and Engineering Mechanics

19. Decision of the University Human Subjects Review Committee:

☐ Project approved  □ Project not approved  □ No action required

Patricia M. Keith
Name of Committee Chairperson  11/22/98
Date  Signature of Committee Chairperson

GC 1/98
Informed Consent to Participate in Research

Department of Aerospace Engineering and Engineering Mechanics
Iowa State University
Ames, IA 50011

You are being asked to volunteer as a participant in a research study. This form is designed to provide you with information about this study and to answer your questions.

1. Title of Research Study
A Three-Dimensional, Dynamic Model of the Human Body During Lifting

2. Project Director
Name: Jason C. Gillette
Address: 0020 Black Engineering
Telephone: (515) 294-2975

3. Purpose of the Research
The purpose of this study is to develop a mathematical model to predict joint forces and moments using experimental data collected during lifting trials.

4. Procedures for this Research
Orientation for Lifting Trials
Prior to the start of the study, there will be an orientation session to familiarize you with the lifting motions to be analyzed and the equipment used during the data collection. At this time you will be able to ask questions and obtain further information about all aspects of the study. You will be asked to perform lifting trials consisting of four repetitions for combinations of two different lifting motions and two different lifting styles, for a total of sixteen trials. The object to be lifted is a crate containing free weights, and you will be allowed to warm up and choose a lifting weight that is comfortable to you. You will be limited to choosing a lifting weight that is equal to or less than fifteen percent of your body weight. One motion involves lifting the crate from ground level in front of the body to waist level in front of the body, and the other motion involves lifting the crate from ground level in front of the body to waist level on the side of the body. The two lifting styles are lifting primarily with the legs and lifting primarily with the back and arms.

Anthropometric Measurements and Marker Placement
During the lifting trials, you will be asked to wear biking shorts or equivalent clothing if you are a male and biking shorts with a jogging bra or equivalent clothing if you are a female. Before performing the lifting trials, a set of anthropometric measurements will be performed to determine the size and shape of your body segments. These measurements will be taken with a standard medical scale, a measuring tape, and a beam caliper. A set of retroreflective markers will be attached to highlight anatomical landmarks on your body. These markers will be attached to your skin by way of double-sided adhesive tape. You will be able to review a list of all measurements before they are taken and a list of marker locations before they are placed.
Data Collection
During the lifting trials, the markers that have been placed on your body will be tracked by four video cameras. An additional conventional video camera will also film you to aid in marker identification during data analysis. You will be standing with each foot on a separate force platform while performing the lifting trials. These platforms will be used to measure the ground reaction forces that occur between your feet and the force platform. Marker tracking by video cameras and force platform measurements are common procedures used in biomechanics. The total time for orientation, measurements, and the lifting trials is expected to be about one and a half hours.

5. Potential Risks or Discomforts

There are minimal risks associated with the lifting trials you will be performing. You will be limited to fifteen percent of your body weight or less in your choice of lifting weight, which is well below the level where injuries would be expected to occur. The amount of weight you will be lifting is similar to what you might experience in a job that required lightweight manual materials handling. You will be allowed to rest between trials if desired. You will be asked if you are uncomfortable with any of the proposed anthropometric measurements or marker placements before they are performed. Some minor irritation might occur upon removal of the markers. There are no invasive procedures used in this study.

6. Potential Benefits to you or Others

There will be no direct benefits to you as a subject in this study. This study may lead to a better method of determining joint forces and moments during lifting through use experimental techniques. It may also lead to a method for injury prevention.

7. Alternate Treatment or Procedures, if Applicable

You have the option of not participating in the study. You are also free to withdraw from the study at any time without consequence.

I understand that will ____/ will not ____ receive money for my participation in this study.

I understand that will ____/ will not ____ be charged expenses for my participation in this study.

I understand that I am free to withdraw my consent and discontinue participation in this research project at any time without prejudice towards me.

Emergency treatment of any injuries that may occur as a direct result of participation in this research will be treated at the Iowa State University Student Health Services, Student Services Building, and/or referred to Mary Greely Medical center or another physician. Compensation for treatment of any injuries that may occur as a direct result of participation in this research may or may not be paid by Iowa State University depending on the Iowa Tort Claims Act. Claims for compensation will be handled by the Iowa State University Vice president for Business and Finance.
My questions on any aspect of this research project are welcomed. At the conclusion of this study I will be informed of the results. My results will be kept confidential and should the data be used in a publication of the results, my name or any identifying characteristics will not be reported.

Signatures

I have fully explained to
the nature and purpose of the above study and the benefits and risks that are involved in participation of the study. I have answered and will answer all questions to the best of my ability.

Signature of Principal Investigator Obtaining Consent

Date

I have been fully informed of the above-described procedure with its possible benefits and risks and I have received a copy of this description. I have given permission for my participation in this study.

Signature of Participant

Date

Signature of Witness

Date
APPENDIX B: MARKER TRACKING PERCENTAGES

An initial set of thirty-two reflective markers was tracked by the video cameras during the lifting trials, but some of these markers were obscured during the motion. Figure B.1 shows the tracking percentages for the symmetric leglifts, Figure B.2 for the asymmetric leglifts, Figure B.3 for the symmetric backlifts, and Figure B.4 for the asymmetric backlifts. After eliminating the poorly tracked markers, a set of twenty-three markers remained and their positions were further analyzed by the 3dLift biomechanical model.
Table B.1: Tracking Percentages for Symmetric Leglifts

<table>
<thead>
<tr>
<th>Marker</th>
<th>Trial 1</th>
<th>Trial 2</th>
<th>Trial 3</th>
<th>Trial 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Right fifth metatarsal</td>
<td>98.4%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Right first metatarsal</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Right lateral malleolus</td>
<td>95.5%</td>
<td>96.2%</td>
<td>88.3%</td>
<td>95.6%</td>
</tr>
<tr>
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<td>71.3%</td>
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</tr>
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</tr>
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* Indicates markers that were eliminated
Table B.2: Tracking Percentages for Asymmetric Leglifts

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<td>81.5%</td>
<td>100.0%</td>
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<td>95.4%</td>
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<td>71.9%</td>
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<td>52.9%</td>
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<td>95.4%</td>
<td>100.0%</td>
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<td>92.9%</td>
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<td>96.2%</td>
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<td>85.8%</td>
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<td>76.7%</td>
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<td>100.0%</td>
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<td>96.2%</td>
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<td>94.4%</td>
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<td>81.0%</td>
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</table>

* Indicates markers that were eliminated, # Indicates trial eliminated
Table B.3: Tracking Percentages for Symmetric Backlifts

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<th>Trial 10</th>
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<th>Trial 12</th>
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<td>100.0%</td>
<td>100.0%</td>
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<td>100.0%</td>
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<td>100.0%</td>
<td>94.4%</td>
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<td>98.4%</td>
</tr>
<tr>
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<td>100.0%</td>
<td>100.0%</td>
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<td>100.0%</td>
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<tr>
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</tr>
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<td>Left greater trochanter</td>
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<td>74.3%</td>
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<td>86.2%</td>
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<tr>
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<td>66.8%</td>
<td>86.2%</td>
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<td>94.4%</td>
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<td>62.9%</td>
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* Indicates markers that were eliminated
Table B.4: Tracking Percentages for Asymmetric Backlifts

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<td>100.0%</td>
<td>100.0%</td>
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<td>100.0%</td>
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<td>100.0%</td>
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<td>94.1%</td>
<td>74.0%</td>
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<td>79.1%</td>
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<td>99.2%</td>
<td>99.7%</td>
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<td>66.2%</td>
<td>95.7%</td>
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<td>77.5%</td>
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<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>78.3%</td>
</tr>
<tr>
<td>Right styline</td>
<td>99.5%</td>
<td>99.5%</td>
<td>100.0%</td>
<td>93.6%</td>
</tr>
<tr>
<td>Right second metacarpel</td>
<td>100.0%</td>
<td>100.0%</td>
<td>98.2%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Right fifth metacarpel*</td>
<td>17.5%</td>
<td>23.3%</td>
<td>66.5%</td>
<td>57.2%</td>
</tr>
<tr>
<td>Left acromion</td>
<td>89.0%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Left radiale</td>
<td>64.8%</td>
<td>91.3%</td>
<td>98.7%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Left styline</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Left fifth metacarpel*</td>
<td>99.5%</td>
<td>92.1%</td>
<td>97.7%</td>
<td>62.0%</td>
</tr>
<tr>
<td>Left second metacarpel</td>
<td>100.0%</td>
<td>98.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Vertex</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
<td>100.0%</td>
</tr>
<tr>
<td>Cervicale*</td>
<td>55.1%</td>
<td>51.5%</td>
<td>42.3%</td>
<td>39.6%</td>
</tr>
<tr>
<td>Right jaw*</td>
<td>31.4%</td>
<td>73.3%</td>
<td>100.0%</td>
<td>77.5%</td>
</tr>
<tr>
<td>Left jaw*</td>
<td>59.1%</td>
<td>54.5%</td>
<td>62.4%</td>
<td>50.3%</td>
</tr>
<tr>
<td>Right medial elbow*</td>
<td>32.4%</td>
<td>16.6%</td>
<td>0.0%</td>
<td>17.4%</td>
</tr>
<tr>
<td>Left medial elbow*</td>
<td>43.1%</td>
<td>56.9%</td>
<td>16.5%</td>
<td>54.3%</td>
</tr>
</tbody>
</table>

* Indicates markers that were eliminated
APPENDIX C: DERIVATION OF ANTHROPOMETRIC EQUATIONS

The anthropometric equations used in the 3dLift model were based on averages derived from previous research studies. This appendix lists the original sources of data that were used to develop these equations, along with any adjustments that were made for differing anthropometric parameters. The combination of these studies appear as the segment mass percentages in Table 3.4, the segment moments of inertia in Tables 3.5 and 3.6, the segment joint centers in Table 5.3, and the segment mass centers in Table 5.4.
Table C.1: Data Sources for Segment Mass Percentages

<table>
<thead>
<tr>
<th>Segment</th>
<th>Source</th>
<th>Original Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot</td>
<td>1</td>
<td>1.47%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>1.37%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>1.5%</td>
</tr>
<tr>
<td>Calf</td>
<td>1</td>
<td>4.35%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>4.33%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>3.7%</td>
</tr>
<tr>
<td>Thigh*¹</td>
<td>1</td>
<td>10.27%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>14.165%</td>
</tr>
<tr>
<td></td>
<td>3*²</td>
<td>13.4%</td>
</tr>
<tr>
<td>Lower Torso</td>
<td>2*³</td>
<td>27.5%</td>
</tr>
<tr>
<td></td>
<td>4*⁴</td>
<td>28.5%</td>
</tr>
<tr>
<td>Upper Torso</td>
<td>2</td>
<td>15.96%</td>
</tr>
<tr>
<td>Upper Arm</td>
<td>1</td>
<td>2.63%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>2.707%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>2.7%</td>
</tr>
<tr>
<td>Forearm</td>
<td>1</td>
<td>1.61%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>1.625%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>1.3%</td>
</tr>
<tr>
<td>Hand</td>
<td>1</td>
<td>0.65%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>0.614%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>0.6%</td>
</tr>
<tr>
<td>Head</td>
<td>1</td>
<td>7.28%</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>6.94%</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>7.3%</td>
</tr>
</tbody>
</table>

Sources:
1 Clauser et al. (1969)
2 Zatsiorsky and Seluyanov (1983)
3 Clarys and Marfell-Jones (1994)
4 Pearsall et al. (1996)

Except where noted, segment masses as a percentage of total body mass were derived by averaging the mass percentages of the sources listed above.

*¹ Thigh segment mass percentages were derived by subtracting the combined mass percentages of the other segments from 100%.

*² Average of right thigh and left thigh.

*³ Sum of lower part of the torso and middle part of the torso.

*⁴ Sum of lower torso, middle torso, and T12 through one-half of T10 level.
Table C.2: Data Sources for Lower Body Segment Moments of Inertia

<table>
<thead>
<tr>
<th>Segment</th>
<th>Source</th>
<th>Original Data</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Foot</strong></td>
<td>1</td>
<td>( I_{ii} = -0.001548 + 0.0000144A_1 + 0.00088A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{ii} = 0.00141A_1 \left( A_2^2 + A_3^2 \right) - 0.0008 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = -0.009709 + 0.0000414A_1 + 0.00614A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = 0.00023A_1 \left( 3A_2^2 + 4A_3^2 \right) + 0.00022 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = -0.01 + 0.000048A_1 + 0.00626A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = 0.00021A_1 \left( 3A_2^2 + 4A_3^2 \right) + 0.00067 )</td>
</tr>
<tr>
<td><strong>Calf</strong></td>
<td>1</td>
<td>( I_{ii} = -0.1105 + 0.000459A_1 + 0.0663A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{ii} = 0.00387A_1 \left( (A_7 - A_5)^2 + 0.076A_8^2 \right) + 0.00138 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = -0.1152 + 0.0004594A_1 + 0.06815A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = 0.00347A_1 \left( (A_7 - A_5)^2 + 0.076A_8^2 \right) + 0.00511 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = -0.00705 + 0.0001134A_1 + 0.003A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = 0.00041A_1A_2^2 + 0.00012 )</td>
</tr>
<tr>
<td><strong>Thigh</strong></td>
<td>1</td>
<td>( I_{ii} = -0.3557 + 0.00317A_1 + 0.1861A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{ii} = 0.070A_1 \left( (A_9 - A_7)^2 + 0.076A_{10}^2 \right) + 0.00161 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = -0.3690 + 0.003202A_1 + 0.1924A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = 0.0762A_1 \left( (A_9 - A_7)^2 + 0.076A_{10}^2 \right) + 0.01153 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = -0.00135 + 0.000113A_1 - 0.0228A_2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = 0.00151A_1A_{10}^2 + 0.00305 )</td>
</tr>
<tr>
<td><strong>Lower Torso</strong></td>
<td>3</td>
<td>( I_{ii} = m_7 \left( (A_{14} - A_9)^2 + 0.75A_{13}^2 \right) / 12 )</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>( I_{jj} = m_7 \left( (A_{14} - A_9)^2 + 0.75A_{16}^2 \right) / 12 )</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>( I_{kk} = m_7 \left( 0.25A_{13}^2 + 0.25A_{16}^2 \right) / 4 )</td>
</tr>
</tbody>
</table>

Sources:
1 Zatsiorsky and Seluyanov (1983)
2 Vaughan et al. (1992)
3 Hanavan (1964)

Except where noted, the segment moment of inertia equations were scaled from the equations of Vaughan et al. (1992). Scaling factors were calculated by putting common anthropometric data from Clauser et al. (1969) into the sources listed above and averaging.

*1 After adjusting according to de Leva (1996a), the Hanavan (1964) model was also used to determine scaling factors.

*2 Scaled from the equations of Hanavan (1964) using the data of Pearsall et al. (1996) to determine scaling factors.
Table C.3: Data Sources for Upper Body Segment Moments of Inertia

<table>
<thead>
<tr>
<th>Segment</th>
<th>Source</th>
<th>Original Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Upper Torso*1</td>
<td>1</td>
<td>( I_{ii} = m_{8} \left( 0.75A_{15}^2 + (A_{18} - A_{14})^2 \right)/12 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = m_{8} \left( 0.75A_{16}^2 + (A_{18} - A_{14})^2 \right)/12 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = m_{8} \left( 0.75A_{15}^2 + 0.25A_{16}^2 \right)/4 )</td>
</tr>
<tr>
<td>Upper Arm*2</td>
<td>2</td>
<td>( I_{ii} = -0.02507 + 0.000156 A_{1} + 0.01512 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = -0.0232 + 0.0001525 A_{1} + 0.01343 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = -0.00169 + 0.0000662 A_{1} + 0.000435 A_{2} )</td>
</tr>
<tr>
<td>Forearm*2</td>
<td>2</td>
<td>( I_{ii} = -0.0064 + 0.000095 A_{1} + 0.0034 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = -0.00679 + 0.0000855 A_{1} + 0.00376 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = 0.000566 + 0.0000306 A_{1} - 0.00088 A_{2} )</td>
</tr>
<tr>
<td>Hand</td>
<td>1</td>
<td>( I_{ii} = m_{11}A_{27}^2 /10 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = m_{11}A_{27}^2 /10 )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = m_{11}A_{27}^2 /10 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = -0.000626 + 0.0000762 A_{1} + 0.000347 A_{2} )</td>
</tr>
<tr>
<td>Head</td>
<td>1</td>
<td>( I_{ii} = 0.2 m_{15} \left( 0.25(A_{2} - A_{18})^2 + A_{31}^2 \right) )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{ii} = -0.0078 + 0.0001171 A_{1} + 0.01519 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{jj} = 0.2 m_{15} \left( 0.25(A_{2} - A_{18})^2 + A_{31}^2 \right) )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{jj} = -0.0112 + 0.000143 A_{1} + 0.0173 A_{2} )</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>( I_{kk} = 0.4 m_{15}A_{31}^2 )</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>( I_{kk} = 0.00616 + 0.000172 A_{1} + 0.000814 A_{2} )</td>
</tr>
</tbody>
</table>

Sources:
1 Hanavan (1964)
2 Zatsiorsky and Seluyanov (1983)

Unless otherwise noted, segment moment of inertia equations were scaled from the model of Hanavan (1964). Scaling factors were calculated by putting common anthropometric data from Clauser et al. (1969) into the sources listed above and averaging.

*1 Scaled from the equations of Hanavan (1964) using the data of Pearsall et al. (1996) to determine scaling factors.

*2 Equations were scaled using a format analogous to the lower body equations of Vaughan et al. (1992) without a constant. After adjusting according to de Leva (1996a) if necessary, the Hanavan (1964) model was used to determine scaling factors.
### Table C.4: Data Sources for Joint Center Positions

<table>
<thead>
<tr>
<th>Joint</th>
<th>Source</th>
<th>Original Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ankle</td>
<td>1</td>
<td>-0.016(A₇ - A₅)</td>
</tr>
<tr>
<td>Knee</td>
<td>1</td>
<td>0.074(A₉ - A₇)</td>
</tr>
<tr>
<td>Hip</td>
<td>1</td>
<td>0.007(A₉ - A₇)</td>
</tr>
<tr>
<td>T10/T11 Disk*¹</td>
<td>2</td>
<td>Center of mass is 4.6 cm anterior to the vertebral centroid at the T10 level</td>
</tr>
<tr>
<td>Shoulder</td>
<td>1</td>
<td>-0.104A₂₁</td>
</tr>
<tr>
<td>Elbow</td>
<td>1</td>
<td>0.043A₂₁</td>
</tr>
<tr>
<td>Wrist</td>
<td>1</td>
<td>0.006A₂₄</td>
</tr>
<tr>
<td>Neck*²</td>
<td>2</td>
<td>Center of mass is 0.8 cm anterior to the vertebral centroid at the T1 level</td>
</tr>
</tbody>
</table>

**Sources:**
1 de Leva (1996b)
2 Pearsall et al. (1996)

Unless otherwise noted, locations of segment joint centers were estimated using de Leva (1996b).

*¹ Using Pearsall et al. (1996), center of mass is approximately 0.3 cm posterior to center of volume at the T10 level. Therefore, vertebral centroid is approximately 4.9 cm posterior to the center of volume at the T10 level. Using average chest depth from Clauser et al. (1969), location of joint center is estimated as percentage of chest depth.

*² Using Pearsall et al. (1996), center of mass is approximately 0.07 cm posterior to center of volume at the T1 level. Therefore, vertebral centroid is approximately 0.87 cm posterior to the center of volume at the T1 level. Using average neck circumference from Clauser et al. (1969), location of joint center is estimated as percentage of neck radius.
Table C.5: Data Sources for Segment Mass Center Positions

<table>
<thead>
<tr>
<th>Segment</th>
<th>Source</th>
<th>Original Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot*†</td>
<td>1</td>
<td>0.4485 A₃</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>0.4415 A₃</td>
</tr>
<tr>
<td>Calf</td>
<td>3*²</td>
<td>0.6177 (A₇ - A₃)</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>0.5800 (A₇ - A₃)</td>
</tr>
<tr>
<td></td>
<td>5*²</td>
<td>0.5945 (A₇ - A₃)</td>
</tr>
<tr>
<td>Thigh</td>
<td>1</td>
<td>0.6281 (A₉ - A₇)</td>
</tr>
<tr>
<td></td>
<td>3*²</td>
<td>0.6337 (A₉ - A₇)</td>
</tr>
<tr>
<td></td>
<td>4</td>
<td>0.6100 (A₉ - A₇)</td>
</tr>
<tr>
<td></td>
<td>5*²</td>
<td>0.6249 (A₉ - A₇)</td>
</tr>
<tr>
<td>Lower Torso</td>
<td>6*³</td>
<td>Center of mass is 16.9 cm below T10 level</td>
</tr>
<tr>
<td></td>
<td>6*⁴</td>
<td>Vertebral centroid is 4.9 cm posterior to center of volume at T10 level</td>
</tr>
<tr>
<td>Upper Torso</td>
<td>6*⁵</td>
<td>Center of mass is 8.75 cm above T10 level</td>
</tr>
<tr>
<td></td>
<td>6*⁶</td>
<td>Vertebral centroid is 4.9 cm posterior to center of volume at T10 level</td>
</tr>
<tr>
<td>Upper Arm</td>
<td>1</td>
<td>0.5130 A₂₁</td>
</tr>
<tr>
<td></td>
<td>3*²</td>
<td>0.5140 A₂₁</td>
</tr>
<tr>
<td>Forearm</td>
<td>1</td>
<td>0.3742 A₂₄</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>0.4274 A₂₄</td>
</tr>
<tr>
<td></td>
<td>3*²</td>
<td>0.3811 A₂₄</td>
</tr>
<tr>
<td></td>
<td>5*²</td>
<td>0.4238 A₂₄</td>
</tr>
<tr>
<td>Hand</td>
<td>1</td>
<td>0.8298 A₂₆</td>
</tr>
<tr>
<td></td>
<td>5*²</td>
<td>0.7957 A₂₆</td>
</tr>
<tr>
<td>Head</td>
<td>2*⁷</td>
<td>0.613 (A₂ - A₁₈)</td>
</tr>
</tbody>
</table>

Sources:
1 Clauser et al. (1969)  4 Vaughan et al. (1992)
2 Zatsiorsky and Seluyanov (1983)  5 de Leva (1996a)
3 Hinrichs (1990)  6 Pearsall et al. (1996)

Unless otherwise noted, the location of the segment mass centers were averaged from the sources listed above.

*¹ \( \phi = \tan^{-1}(A₃ / (A₃ - A₂₄)) \) takes into account the tilt of the foot from ankle to ground.

*² Adjusted using data of de Leva (1996b).

*³ Location of segment mass center found by dividing by length from S5 to T10.

*⁴ Location of segment mass center is approximately 0.9 cm posterior to the center of volume. Ratio formed by dividing by average chest depth from Clauser et al. (1969).

*⁵ Location of segment mass center found by dividing by length from T1 to T10.

*⁶ Location of segment mass center is approximately 0.1 cm posterior to the center of volume. Ratio formed by dividing by average chest depth from Clauser et al. (1969).

*⁷ Adjusted using data of de Leva (1996a).
APPENDIX D: SIGNAL PROCESSING SUBROUTINES

Appendix D contains the Subroutine Descent, Function Sumsquares, and Subroutine Butterworth from the 3dLift FORTRAN program. Subroutine Descent used the method of steepest descent to improve initial guesses for obscured marker extrapolation and called Function Sumsquares. Subroutine Butterworth implemented a fourth-order Butterworth low-pass filter to attenuate data noise above a selected cutoff frequency.
SUBROUTINE DESCENT(M1,M2,M3,M4,MARKER,L2,L3,L4,LENGTH,N,X1)

* Subroutine Descent uses the method of steepest descent (Burden and Faires, 1989) to improve an initial guess for extrapolation of obscured marker data. It is called by Subroutine Extrapolation, which is in turn called by the main program of 3dLift.

REAL A0, A1, A2, A3, F(3), G0, G1, G2, G3, H1, H2, H3, INIT(3)
REAL J(3,3), LENGTH(35), MARKER(430,23,3), MINIMAL(2), TOL, X0(3)
REAL X1(3), X2(3), X3(3), Z(3), Z0
INTEGER ITER, K, L2, L3, L4, M1, M2, M3, M4, N

* A0 to A3 – Define search interval
* F(3) – The lengths between visible markers and the obscured marker
* G0 to G3 – Minimization function value, which is the sum of squares of the lengths between markers
* H1 to H3 – Quadratic interpolation constants over search interval to determine critical search point
* INIT(3) – Initial guesses for obscured marker positions
* ITER – Maximum number of iterations
* J(3,3) – Jacobian matrix of the minimization function
* K – Number of iterations
* L2 to L4 – Lengths between surrounding visible markers and obscured marker
* LENGTH(35) – Lengths between markers from anatomical position data
* M1 to M4 – Obscured marker number and visible surrounding marker numbers
* MARKER(430,23,3) – Marker position data from video tracking
* MINIMAL(2) – Minimal value for minimization function and its point in search interval
* N – Data point at which marker is obscured
* TOL – Error tolerance
* X0(3) to X3(3) – Estimated positions of obscured marker
* Z(3) – Normalized gradient of minimization function
* Z0 – Magnitude of gradient

* Determines minimization function, Jacobian matrix, and gradient vector for initial guess of obscured marker position.

INIT(1) = X1(1)
INIT(2) = X1(2)
INIT(3) = X1(3)
TOL = 1.0E-6
ITER = 100
K = 0

10 IF (K .LT. ITER) THEN
   G1 = SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X1)
   F(1) = (X1(1) - MARKER(N,M2,1))**2 + (X1(2) -
   * MARKER(N,M2,2))**2 + (X1(3) - MARKER(N,M2,3))**2 -
   * LENGTH(L2)**2
   F(2) = (X1(1) - MARKER(N,M3,1))**2 + (X1(2) -
   * MARKER(N,M3,2))**2 + (X1(3) - MARKER(N,M3,3))**2 -
   * LENGTH(L3)**2
   F(3) = (X1(1) - MARKER(N,M4,1))**2 + (X1(2) -

   IF (G1 .GT. MINIMAL(1)) THEN
      INIT(1) = X1(1)
      INIT(2) = X1(2)
      INIT(3) = X1(3)
      TOL = 1.0E-6
      ITER = 100
      K = 0
   END IF

END
* MARKER(N,M4,2)**2 + (X1(3) - MARKER(N,M4,3))**2 -
* LENGTH(L4)**2
J(1,1) = 2.0*(X1(1) - MARKER(N,M2,1))
J(1,2) = 2.0*(X1(2) - MARKER(N,M2,2))
J(1,3) = 2.0*(X1(3) - MARKER(N,M2,3))
J(2,1) = 2.0*(X1(1) - MARKER(N,M3,1))
J(2,2) = 2.0*(X1(2) - MARKER(N,M3,2))
J(2,3) = 2.0*(X1(3) - MARKER(N,M3,3))
J(3,1) = 2.0*(X1(1) - MARKER(N,M4,1))
J(3,2) = 2.0*(X1(2) - MARKER(N,M4,2))
J(3,3) = 2.0*(X1(3) - MARKER(N,M4,3))
Z(1) = 2.0*F(1)*J(1,1) + 2.0*F(2)*J(2,1) + 2.0*F(3)*J(3,1)
Z(2) = 2.0*F(1)*J(1,2) + 2.0*F(2)*J(2,2) + 2.0*F(3)*J(3,2)
Z(3) = 2.0*F(1)*J(1,3) + 2.0*F(2)*J(2,3) + 2.0*F(3)*J(3,3)
Z0 = SQRT(Z(1)**2 + Z(2)**2 + Z(3)**2)

*********************************************************************************************************************
* Normalizes gradient vector and uses this information to determine first estimation of the obscured marker
* position beyond the initial guess
*********************************************************************************************************************

IF (Z0.EQ.0.0) THEN
  WRITE(*,*) 'Error: Zero gradient in extrapolation'
  STOP
END IF
Z(1) = Z(1)/Z0
Z(2) = Z(2)/Z0
Z(3) = Z(3)/Z0
A1 = 0.0
MINIMAL(1) = A1
MINIMAL(2) = G1
A3 = 1.0
X3(1) = X1(1) - A3*Z(1)
X3(2) = X1(2) - A3*Z(2)
X3(3) = X1(3) - A3*Z(3)
G3 = SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X3)

*********************************************************************************************************************
* Determines whether or not estimation of the obscured marker position can be improved by cutting the search
* interval in half.
*********************************************************************************************************************

20 IF (G3.GE.G1) THEN
  A3 = A3/2.0
  IF (A3.LT.TOL/2.0) THEN
    MARKER(N,M1,1) = X1(1)
    MARKER(N,M2,2) = X1(2)
    MARKER(N,M3,3) = X1(3)
    GOTO 30
  END IF
  X3(1) = X1(1) - A3*Z(1)
  X3(2) = X1(2) - A3*Z(2)
  X3(3) = X1(3) - A3*Z(3)
  G3 = SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X3)
GOTO 20
END IF
MINIMAL(1) = A3
MINIMAL(2) = G3
A2 = A3/2.0
X2(1) = X1(1) - A2*Z(1)
X2(2) = X1(2) - A2*Z(2)
X2(3) = X1(3) - A2*Z(3)
G2 = SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X2)
IF (G2 .LT. G3) THEN
  MINIMAL(1) = A2
  MINIMAL(2) = G2
END IF

* Uses quadratic interpolation to determine a critical point in the search interval to check for improvement in
* the estimation of the obscured marker position.

H1 = (G2 - G1)/A2
H2 = (G3 - G2)/(A3 - A2)
H3 = (H2 - H1)/A3
A0 = 0.5*(A2 - H1/H3)
X0(1) = X1(1) - A0*Z(1)
X0(2) = X1(2) - A0*Z(2)
X0(3) = X1(3) - A0*Z(3)
G0 = SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X0)
IF (G0 .LT. MINIMAL(2)) THEN
  MINIMAL(1) = A0
  MINIMAL(2) = G0
END IF

* Adjusts initial guess of the obscured marker position to a new estimation of the marker position that results in
* the lowest value of the minimization function. The initial guess and the new estimated position are then
* combined to form the extrapolated marker position.

X1(1) = X1(1) - MINIMAL(1)*Z(1)
X1(2) = X1(2) - MINIMAL(1)*Z(2)
X1(3) = X1(3) - MINIMAL(1)*Z(3)
IF (ABS(MINIMAL(2) - G1) .LT. TOL) THEN
  MARKER(N,M1,1) = (1.0*X1(1) + 3.0*INIT(1))/4.0
  MARKER(N,M1,2) = (1.0*X1(2) + 3.0*INIT(2))/4.0
  MARKER(N,M1,3) = (1.0*X1(3) + 3.0*INIT(3))/4.0
GOTO 30
END IF
K = K + 1
GOTO 10
END IF
WRITE(*,*).'Error: Iterations exceeded in extrapolation'

30 END
FUNCTION SUMSQUARES(L2,L3,L4,LENGTH,M2,M3,M4,MARKER,N,X)

* Determines the value of the minimization function used to estimate obscured marker positions when 
  * extrapolation is necessary. Function Sumsquares is called by Subroutine Descent, which is called by 
  * Subroutine Extrapolation, which is in turn called by the main program of 3dLift.

REAL F(3), LENGTH(35), MARKER(430,23,3), SUMSQUARES, X(3)
INTEGER L2, L3, L4, M2, M3, M4, N

* F(3) - The lengths between visible markers and the obscured marker 
* L2 to L4 - Lengths between surrounding visible markers and obscured marker 
* LENGTH(35) – Lengths between markers from anatomical position data 
* M2 to M4 – Visible surrounding marker numbers 
* MARKER(430,23,3) – Marker position data from video tracking 
* N – Data point at which marker is obscured 
* TOL – Error tolerance 
* SUMSQUARES – Value of minimization function 
* X(3) – Estimated position of obscured marker 

* Determines the minimization function value by finding the lengths between the visible markers and the 
  * obscured marker and summing the squares of these lengths.

F(1) = (X(1) - MARKER(N,M2,1))**2 + (X(2) - 
  * MARKER(N,M2,2))**2 + (X(3) - MARKER(N,M2,3))**2 - 
  * LENGTH(L2)**2 
F(2) = (X(1) - MARKER(N,M3,1))**2 + (X(2) - 
  * MARKER(N,M3,2))**2 + (X(3) - MARKER(N,M3,3))**2 - 
  * LENGTH(L3)**2 
F(3) = (X(1) - MARKER(N,M4,1))**2 + (X(2) - 
  * MARKER(N,M4,2))**2 + (X(3) - MARKER(N,M4,3))**2 - 
  * LENGTH(L4)**2 
SUMSQUARES = F(1)**2 + F(2)**2 + F(3)**2

END
SUBROUTINE BUTTERWORTH(BACK, FC, FRAME, K, RAW, S)

* Attenuates noise above a selected cutoff frequency by running data through a fourth-order, low-pass
* Butterworth filter (Winter, 1990). Subroutine Butterworth is called by Subroutines Filter, AngVelocity,
* AngAccel, LinAccel, and ForcePlate.

INTEGER FRAME, I, K, N, S
REAL A0(3), A1(3), A2(3), B1(3), B2(3), BACK(430, K, 3), FC(3)
REAL FCA(3), FOR(430, K, 3), FS, K1(3), K2(3), K3(3), PI
REAL RAW(430, K, 3), WCA(3)

* A0(3) to A2(3) – Non-recursive Butterworth filter coefficients
* B1(3) to B2(3) – Recursive Butterworth filter coefficients
* BACK(430, K, 3) – Data after it has been run a second time backwards through the Butterworth filter
* FC(3) – Cutoff frequency
* FCA(3) – Cutoff frequency adjusted for two passes through the Butterworth filter
* FOR(430, K, 3) – Data after it has been run a first time forward through the Butterworth filter
* FRAME – Number of data points in trial
* FS – Sampling frequency
* I – Axes i, j, k
* K – Number of markers, segments, or platforms in array
* K1(3) to K3(3) – Parameters used to determine Butterworth filter coefficients
* N – Data point being filtered
* PI – Numerical constant Pi
* RAW(430, K, 3) – Raw, unfiltered data
* S – Marker number, segment number, or platform number of data to be filtered

FS = 120.0
PI = 4.0 * ATAN(1.0)
DO 10 I = 1, 3
   FCA(I) = (FS/PI) * ATAN(TAN(PI*FC(I)/FS)/((2.0**0.5 - 1.0)**0.25))
   WCA(I) = TAN(PI*FCA(I)/FS)
   K1(I) = 2.0**0.5*WCA(I)
   K2(I) = WCA(I)**2
   A0(I) = K2(I)/(1.0 + K1(I) + K2(I))
   A1(I) = 2.0*A0(I)
   A2(I) = A0(I)
   K3(I) = 2.0*A0(I)/K2(I)
   B1(I) = -2.0*A0(I) + K3(I)
   B2(I) = 1.0 - 2.0*A0(I) - K3(I)
10 CONTINUE
* Leaves first raw data point unchanged, performs three-point moving average on next two points, and runs remaining data forward through the Butterworth filter.

```
DO 20 I = 1, 3
   FOR(1,S,I) = RAW(1,S,I)
   FOR(2,S,I) = (RAW(1,S,I) + 2.0*RAW(2,S,I) + RAW(3,S,I))/4.0
   FOR(3,S,I) = (RAW(2,S,I) + 2.0*RAW(3,S,I) + RAW(4,S,I))/4.0
20 CONTINUE

DO 40 N = 4, FRAME
   DO 30 I = 1, 3
      FOR(N,S,I) = A0(I)*RAW(N,S,I) + A1(I)*RAW(N-1,S,I) +
      *       A2(I)*RAW(N-2,S,I) + B1(I)*FOR(N-1,S,I) +
      *       B2(I)*FOR(N-2,S,I)
   30 CONTINUE
40 CONTINUE
```

* Leaves last data point filtered once, performs three-point moving average on previous two points, and runs remaining data backwards through the Butterworth filter.

```
DO 50 I = 1, 3
   BACK(FRAME,S,D) = FOR(FRAME,S,I)
   BACK(FRAME-1,S,D) = (FOR(FRAME,S,I) + 2.0*FOR(FRAME-1,S,I) +
   *      FOR(FRAME-2,S,I))/4.0
   BACK(FRAME-2,S,D) = (FOR(FRAME-1,S,I) + 2.0*FOR(FRAME-2,S,I) +
   *      FOR(FRAME-3,S,I))/4.0
50 CONTINUE

DO 70 N = FRAME-3, 1, -1
   DO 60 I = 1, 3
      BACK(N,S,D) = A0(I)*FOR(N,S,I) + A1(I)*FOR(N+1,S,I) +
      *       A2(I)*FOR(N+2,S,I) + B1(I)*BACK(N+1,S,I) +
      *       B2(I)*BACK(N+2,S,I)
   60 CONTINUE
70 CONTINUE
```

END
APPENDIX E: CARDAN ANGLES DURING LIFTING

Appendix E shows the Cardan angles of selected segments that were determined by the 3dLift model for the lifting trials. Figure E.1 displays the thigh flexion-extension Cardan angles, Figure E.2 the torso flexion-extension Cardan angles, and Figure E.3 the torso axial rotation Cardan angles. The results have been averaged over four symmetric leglift trials, three asymmetric leglift trials, four symmetric backlift trials, and four asymmetric backlift trials.
Figure E.1: Thigh Flexion-Extension Angles
Figure E.2: Torso Flexion-Extension Angles
Figure E.3: Torso Axial Rotation Angles
APPENDIX F: SELECTED MASS CENTER ACCELERATIONS

Appendix F displays the mass center accelerations of selected segments that were calculated by the 3dLift model. Figure F.1 shows the calf mass center accelerations, Figure F.2 shows the thigh mass center accelerations, and Figure F.3 shows the upper arm mass center accelerations. These results compare the differences between the symmetric and asymmetric leglift trials and the differences between the symmetric and asymmetric backlift trials. The results have been averaged over four symmetric leglift trials, three asymmetric leglift trials, four symmetric backlift trials, and four asymmetric backlift trials.
Figure F.1: Calf Mass Center Accelerations
Figure F.2: Thigh Mass Center Accelerations
Figure F.3: Upper Arm Mass Center Accelerations
Appendix G shows the joint forces and joint moments of selected segments that were calculated by the 3dLift model. Figure G.1 displays the ankle joint forces, Figure G.2 displays the ankle joint moments, Figure G.3 displays the wrist joint forces, and Figure G.4 displays the wrist joint moments. These figures illustrate how the results changed at the ankle and wrist depending on the type of lifting motion. The results have been averaged over four symmetric leglift trials, three asymmetric leglift trials, four symmetric backlift trials, and four asymmetric backlift trials.
Figure G.1: Ankle Joint Forces
Figure G.2: Ankle Joint Moments
Figure G.3: Wrist Joint Forces
Figure G.4: Wrist Joint Moments
REFERENCES


