ABSTRACT

The Influence of Load on Kinematics of Computer-Simulated Sagittal-Plane Lifting

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Researchers have shown that lifting kinematics change predictably with increased load. To test whether these kinematics patterns are intrinsic or voluntary, a computer model was developed to simulate lifting in the sagittal plane. The eight-degree-of-freedom model included the ankle, knee, hip, shoulder, elbow, neck, and two back joints. Strength limits were assigned to model joints according to position-dependent average male data obtained from the literature. Using both forward and inverse dynamics approaches, the model was programmed to lift various loads while tracking lift kinematics measured from a human subject. Simulation results suggest that, contrary to common hypotheses, observed lifting patterns are not dictated by physical law (intrinsic) but are chosen for efficiency and stability (voluntary).

In this study, a method for isolating kinematic dependencies is introduced. It is anticipated that the results will help in the understanding of motion perception, lifting technique, and low-back pain.

The Influence of Load on Kinematics of Computer-Simulated Sagittal-Plane Lifting

by

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LIST OF ABBREVIATIONS

BOS	base of support
COM	center of mass
COP	center of pressure
DOF	degree of freedom
KSD	kinematics specify dynamics
LBP	low-back pain
MAWL	maximum acceptable weight of lift
MMH	manual materials handling
MUR	muscular utilization ratio
vLaH	virtual light actual heavy
vLaL	virtual light actual light
vLaM	virtual light actual medium
vMaH	virtual medium actual heavy
vMaL	virtual medium actual light
vMaM	virtual medium actual medium
vHaH	virtual heavy actual heavy
vHaL	virtual heavy actual light
vHaM	virtual heavy actual medium

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CHAPTER ONE

Introduction

Low-back Pain

In the industrialized world, tons of materials are handled manually each day during the production, transportation, storage, sale, and maintenance of raw and finished goods [1]. As manual materials handling (MMH) has rapidly increased in recent years, so have related incidences of low-back pain (LBP). For instance, the rate of LPB-disability in the US increased 14 times faster than the population rate between 1960 and 1980 [2]. An estimated 70-85% of all US people experience LBP at some time in life [3]. LBP is a disastrous problem both in terms of the prolonged pain that LBP victims must endure and in the economic costs for workers and their employers. Records indicate that work related back injury is the most expensive medical problem in industry [4-6]. Webster and Snook estimated that LBP-related workers' compensation claims directly incurred at least $11.4 billion in 1989. Each year, approximately 2% of the US work force are compensated for back injuries [3], and only 50% of those workers whose leave of absence lasts for 6 months or more eventually return to work [7]. In 1988, one consequence of back and spine impairments was 185 million days of restricted activity in the US [3]. Clearly the repercussions of LBP in manual materials handling are overwhelming and unacceptable.

Lifting and Technique

According to the National Safety Council, in MMH tasks over 50% of work-related low-back injuries are caused by manual lifting as noted by Khalaf et al. [6]. In many cases LBP may be a direct result of the lifting technique adopted. Since it may be more effective to prevent LBP than to cure it, many studies have examined the possibility of prescribing a particular lifting technique in hope of reducing the likelihood of injury. Garg and Herrin explain, "Conventional wisdom of the last forty years has taught us that heavy loads should be lifted with a squat (straight back, bent-knee) posture. The basis for this recommendation has been that it shifts the stresses on the body from the lower back to the legs. According to principles of leverage, it was reasoned, the force imposed upon the lumbar vertebrae increases with the angulation of the trunk [8]."

However, there seem to be little, if any, concrete scientific evidence to support this recommendation, and many researchers have criticized the squat lifting posture [4, 9]. Many have suggested that the stoop technique (relatively straight knee, bent-back) posture may be superior in terms of balance, knee clearance, fatigue, and metabolic energy expended [4, 8, 10, 11]. Furthermore, a growing number of authors believe that the lifting technique spontaneously adopted by the subject may be the least likely to lead to injury [12]. The advantages of the freestyle lift are described by Heiss et al.:

> Persons who are allowed to use a freestyle lift (ie, the technique that is most comfortable for them) adopt the semisquat technique. The semisquat technique has several advantages over the squat technique. The starting position of the semisquat lift raises the body center of mass (COM) and reduces the work of lifting and the rate of energy expenditure. When the knees are more extended, the horizontal distance between the body COM and the knee joint is reduced, thereby reducing the required knee extensor moment and the demands on the quadriceps. Subjects who perform repetitive lifts by using the squat technique transition to the semisquat position as the weight of the load increases or as the leg muscles fatigue. [11]

Many studies have concluded that although the description of initial lifting posture, such as the squat or stoop, provides helpful information about the type of lift employed, the terms are grossly inadequate for the understanding of the coordination involved in the lift [9, 12-18]. Burgess-Limerick et al. reported that “a description of the interjoint coordination and how this coordination is altered by changes in the task may well be equally – or even more – important in describing the lifting techniques [12].” A thorough understanding of lifting coordination is essential if lifting techniques are to be either prescribed or examined to minimize the possibility of injury.

Various methods for analyzing interjoint coordination (e.g., position time series, velocity time series, phase plane, relative phase, and angle-angle plots) facilitate quantitative observation of the kinematics involved in lifting technique. These analysis methods are often used to examine kinematic differences in lifting as task parameters are varied. In particular, several studies have documented consistent changes in interjoint patterns as the load being lifted is increased [12, 15-22]. For example, Scholz reported that as subjects lifted heavier weights, back extension tended to occur increasingly later in the movement than knee extension [16-18].

Studies investigating the impact of load on lifting coordination have raised pertinent questions for researchers who seek to prevent LBP by prescribing safer lifting techniques. Is it physically possible for a lifter to adhere to a prescribed technique under varying conditions of load magnitude? If so, should the load to be lifted be an important factor in the decision of which technique to prescribe? It is clear that by altering lifting technique, the body will experience different loads and forces, which may in turn cause damage that might not have occurred if another lifting strategy had been incorporated.

Therefore, it would be quite beneficial to develop a method for isolating kinematic dependencies related to load so that “the natural, intrinsic (i.e., independent of intentional influences) dynamics of the coordination [17]” could be separated from the voluntary efforts of the lifter.

Perception

The study of perception is another area of research that could benefit from the techniques that seek to isolate intrinsic from intentional dynamics of lifting as load increases. Many perception investigations have revealed that people somehow recognize complex motion patterns from point light displays, which can be made by filming actors with small light sources attached to their major joints. Then the contrast can be adjusted until only the point light sources are visible on film [23]. An example point light display for lifting is illustrated in Figure 1.1. Numerous studies have used point light displays to examine human perception of lifting [24-30]. These researchers concur that lifting technique is consistently affected by load. In perhaps the most important and controversial study on human perception of lifting, Runeson and Frykholm demonstrated that when observers were shown a point light display of an actor pretending to lift a different weight than was actually lifted, viewers discerned both the actual and intended weight [26, 27]. Based on these results, Runeson and Frykholm concluded that there was information present in the kinematics alone that allowed observers to directly infer dynamic properties (i.e., the load). The authors then theorized the kinematics specify dynamics (KSD) principles, which state that “one cannot be deceived by lifted weight because (a) weight is directly perceived through changes in the kinematic pattern and (b) lifters are unable to manipulate all of the kinematics required to match those generated by a different weight [29].” According to the hypotheses, an actor cannot fake the load-dependent intrinsic dynamics of the lifting movement. However, there is also considerable controversy regarding the validity of the KSD principles [17, 25, 26, 27, 28, 29, 31].

Figure 1.1. Point light display frame used for current study's examination of lifting.

One major aim of the current study was to attempt to isolate potential inherent kinematic differences between lifting movements as load is altered. This accomplishment might give evidence to support or contradict the KSD principles proposed by Runeson and Frykholm [26-28].

Modeling and Simulation

Since people lack the dexterity to perfectly manipulate all the kinematics in order to exactly match a particular lifting movement, it is not feasible to isolate kinematic dependencies by purely experimental methods. Therefore, modeling is often more appropriate.

As computational power has increased tremendously, the popularity of modeling and simulation of lifting has grown and is now widely accepted. Biomechanical models and computer simulations of lifting have been used for many situations, including estimation of compressive loads on joints in the lower back [5, 11, 32, 33], evaluation of optimal motion patterns of the lifting movement [34-37], quantification of postural changes [13], and establishment of safe weight limits for people to lift [1, 38].

Computer-simulated models normally predict movements via integration [4, 34]. Initial position and velocity states are supplied to the model, and appropriate joint torques or muscle forces are applied. The model then uses forward dynamics analysis to solve equations of motion for resulting angular accelerations about the model's joints. The acceleration and velocity states are then numerically integrated to provide subsequent states so that the process can be repeated for the next step in time. The result of this integration process is that, by supplying correct initial positions and instantaneous joint torques or forces, the model is able to use the laws of physics to predict the desired motion.

In many cases, the correct instantaneous torques or forces are unknown. Typically, an optimization-based approach must be used to choose various torques or muscular forces that can be applied to the model to bring about a particular motion. The optimization routine minimizes an objective function based on performance criteria that are hypothesized to represent a presumed optimal strategy to accomplish the movement goal [39]. Many techniques for implementing these types of optimizations have been utilized such as simulated annealing [39], computed muscle control [40], and particle swarm [41, 42].

Overview

As individuals pick up heavier loads, they tend to alter their lifting technique and exhibit consistent movement patterns [12, 15-22]. There has been some controversy concerning whether these kinematic patterns are voluntarily adopted or if they are unavoidable due to inherent physical limitations [17, 25-29]. The purpose of this investigation was to determine if physical laws and other factors, such as stability and strength, necessitate differences in lifting kinematics as the magnitude of the load is varied. A computer model was developed to isolate any potential intrinsic (i.e., independent of intentional influences) dynamics that might exist due to varying the load. To this investigator's knowledge, no previous study has been conducted to identify and separate intrinsic from voluntary differences in kinematics for varying loads. It is anticipated that the results will increase understanding of motion perception, lifting technique, and low back pain.

In the current study, an eight-degree-of-freedom biomechanical computer model was constructed to analyze lifting in the sagittal plane. The model was defined to represent an actual subject, who performed experimental lifting trials of three loads: light, medium, and heavy. The subject's experimental lifting kinematics were then used as inputs for the model. The model was constrained to obey the laws of physics, maintain balance, and comply with documented angle-dependent strength limitations of each joint. By performing simulations in which the model was given a particular load and was burdened with generating the same kinematics that the subject displayed as he lifted a different weight, differences between the model's kinematics and those of the experimental trials could be observed to reveal potential movement changes that cannot be avoided as the load is varied. Further analysis was then performed to investigate factors thought to potentially impact lifting kinematics such as joint strength, stability, and energy efficiency.

Chapter 2 contains the methods of collecting experimental data from a single male subject and the corresponding data reduction. In Chapter 3, the development and description of the model are explained. Simulation implementation and its related applications for the current study of lifting are defined in Chapter 4. Experimental results are provided in Chapter 5. In Chapter 6, the results are interpreted, conclusions are drawn, and the significance and limitations of the study are noted. Chapter 7 includes a summary of significant findings and conclusions drawn from the investigation.

CHAPTER TWO

Methods – Collection and Reduction of Experimental Data

Many studies have reported that lifters exhibit common trends to adjust technique in response to load magnitude [12, 15-22]. Since the primary purpose of this study was to investigate these phenomena and provide insight concerning possible causes of these documented patterns, experimental lifting data was collected and analyzed. Kinematics were examined for a single male subject who performed multiple lifting trials of various weights. Of all the recorded trials, one was selected for each load amount to serve as a representative lift for the general lifting population. Representative trials were chosen to fit the most commonly documented kinematic trends that result from changes in load.

Chapter Two Summary

The subject is described in the *Participating Subject* section. In *Lifting Protocol*, instructions to the lifter and the general setup of the experimental procedure are provided. The next section, *Weight Determination*, explains the process that was used to assess the lifting strength potential of the subject. The selection of load amounts for the subject to lift is also justified in this section. The methods used to obtain lifting motion data from the subject are described in *Experimental Lifts*. The techniques used to process the experimental data and the specific analysis methods used to determine the representative experimental trials are presented in the *Data Reduction* section.

Participating Subject

A healthy 23 year-old male subject was recruited from the student population of Baylor University to participate in this study. He reported no injuries, history of low-back disorders, or other motor impairments. The subject signed an informed consent form, approved by the Baylor Institutional Review Board. For reasons discussed in Chapter 6, the study was based entirely on experimental data from this one subject.

Anthropometric measurements were obtained from the subject. His body mass was 91.9 kg, and his height was 1.79 m. Table 2.1 provides measurements of the subject's segment lengths and defines segment endpoints by anatomical reference points on the body [43].

Table 2.1. Subject's anthropometric measurements.

Segment	Endpoints (proximal to distal)	Segment Length (m)
Shank	Knee to ankle center	0.42
Thigh	Hip to knee center	0.38
Pelvis	L4-L5 to trochanter	0.16
Abdomen	T12-L1 to L4-L5	0.11
Thorax	C7-T1 to T12-L1	0.40
Upper arm	Glenohumeral joint to elbow center	0.28
Forearm	Elbow to wrist center	0.35
Head	C7-T1 to ear canal	0.11
Ankle to toe	Ankle center to end toe II	0.21
Ankle to heel	Ankle center to end of heel	0.05

Lifting Protocol

The subject was asked to perform manual lifts of a barbell. The lifting strategy was freestyle (i.e., the subject could choose any lifting technique and speed he desired) rather than specifying a squat technique or some other strategy because, as Scholz pointed out, "allowing subjects to lift in an unconstrained fashion . . . might better reveal the natural, intrinsic (i.e. independent of intentional influences) dynamics of the coordination of this task [17].”

A weight rack was constructed to provide consistent beginning and ending positions for the barbell. Notches were placed on the weight rack to prevent horizontal movement of the barbell. The starting position of the barbell was 0.3 m above the ground, and the ending position was 1.21 m high (i.e., mid-stomach level of standing subject). The rack was also constructed to accommodate a force platform for the subject to stand on.

A six-channel force platform (Bertec Corp., Model 4060-08, Columbus, OH) was used to measure ground reaction forces according to force and moment data in the vertical, horizontal, and lateral directions throughout the lifting movement. The height of the force platform was 0.08 m.

A specific lifting protocol was adopted in order to provide consistency among the lifts and to minimize errors related to modeling assumptions. The subject began each lift in a self-chosen crouched position on the force plate with his shoes off. He was then required to grasp the barbell with his palms facing up. The lift consisted of raising the weight from the beginning position on the rack to the ending position. The subject was instructed not to overexert himself. He was also given a few minutes rest between each lift and was allowed more time if desired. While the movements were to be a self-selected freestyle lifts, the subject was still asked to obey the following conditions:

(1) Once placed, his feet could not move during the lift.

(2) All movement was to be in the sagittal plane (i.e., no abduction or adduction of the arms or lateral movement of the knees).

(3) All movement was to be symmetrical (i.e., arms inline with each other).

(4) The barbell had to be placed gently on the rack.

Weight Determination

The subject participated in a psychophysical experimental session to determine the maximum acceptable weight of lift (MAWL) [44]. In this session, the subject followed the lifting protocol described above while lifting increasingly heavy barbells. The subject's MAWL was determined to be the heaviest load the subject could safely lift in a controlled fashion without feeling overexerted. Several minutes rest was given between each lift to avoid fatigue.

The MAWL was used to select appropriate light, medium, and heavy loads to be lifted in the experimental data collection session. The magnitudes of the three loads were chosen to both (1) avoid over-fatigue of the subject and, (2) select significantly distributed weights so that inherent kinematic differences due to load might be exaggerated and more easily identified. The mass of the light, medium, and heavy loads as well as their relation to the MAWL is provided in Table 2.2.

Table 2.2. Magnitudes of loads lifted, relative to maximum acceptable weight of lift (MAWL) for the subject.

Property	Light	Medium	Heavy	MAWL
Mass (kg)	8.3	22.2	35.9	45.4
Weight (lbs)	18.25	49.00	79.25	105.00
Approximate % of MAWL	20	50	80	100

Experimental Lifts

Two days after the MAWL was determined, the subject participated in an experimental data collection session in which lifting motion data was obtained. The subject was required to follow the same lifting protocol as described above. The subject was allowed as many practice lifts as he desired for each of the three weights so that he could familiarize himself with the lifting process and determine the most comfortable foot placement position. This position was then marked on the force plate to ensure consistent positioning. The light weight was used for most of the practice trials to minimize fatigue. Aside from the practice trials, three lifts were recorded for each load category. Again, fatigue was minimized by specifying recorded trials to be lifted in the order: light, medium, and then heavy. At least four minutes of rest was given between each recorded trial.

The lifting motion data of the recorded trials was obtained via a video motion analysis system (zFlo, Inc., Quincy, MA) consisting of four 60 Hz video cameras. Adhesive reflective markers were attached to the skin to indicate the location of the subject’s ankle, knee, hip, shoulder, elbow, forearm, ear, forehead, and three spine joints. The spine joints were located at L4-L5, T12-L1, and C7-T1, which corresponded to the articulations between the fourth and fifth lumbar vertebrae, the twelfth thoracic and first lumbar vertebrae, and the seventh cervical and first thoracic vertebrae, respectively. Additional markers were attached to the right and left sides of the barbell to verify that the weight was lifted symmetrically. A software package (Simi Reality Motion Systems gmbh, Unterschleissheim, Germany, SIMI°Motion ver. 6.5) was used to collect joint displacement data. The setup of the experimental data collection is illustrated in Figure 2.1.

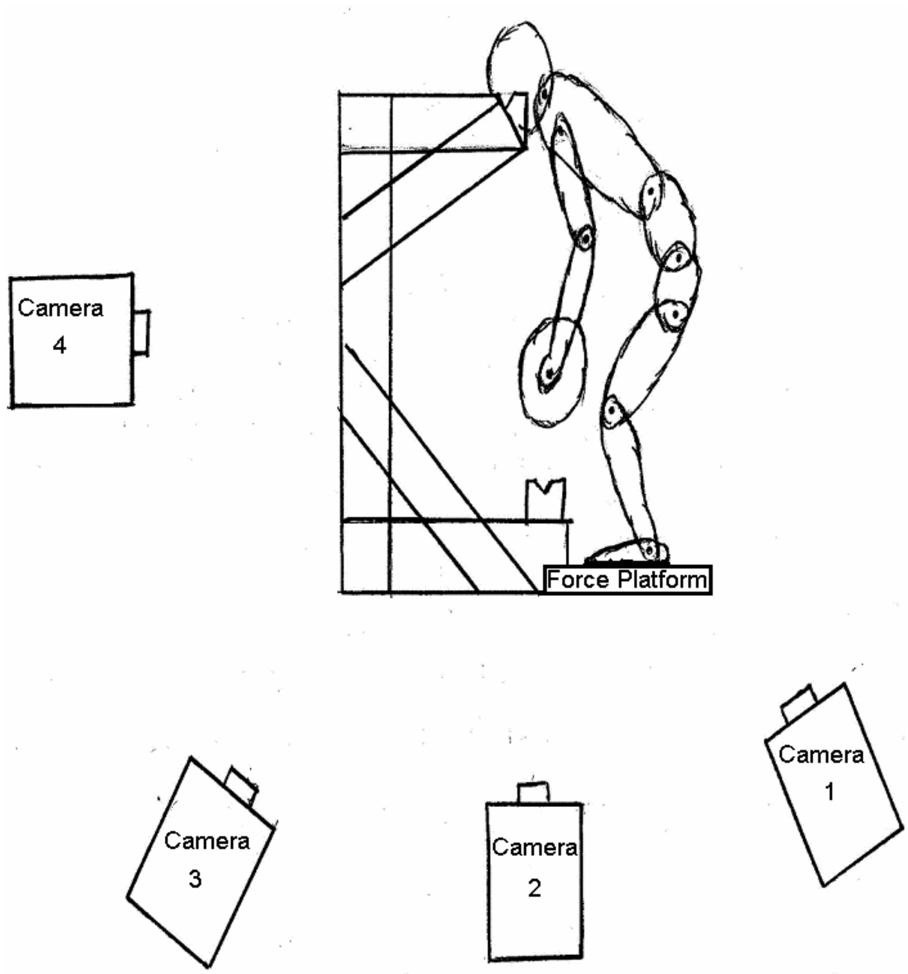


Figure 2.1. Experimental data collection setup. Subject lifts a barbell onto a rack. His movements are recorded by a motion capture system, and a force platform measures ground reaction forces.

Data Reduction

Since the markers for the L4-L5, T12-L1, and C7-T1 joints were placed on the back rather than on the joint centers of rotation, data processing was required to represent the true positions of the spine joints. The SIMI°Motion software package exported the coordinate locations of each reflective marker to a spreadsheet (Microsoft, Redmond, WA, Excel 2002 ver. 10.6501.6735). The Visible Human Male image set [45] was used to determine the anterior/posterior distances from the back joint centers to the skin surfaces on the back. It seemed reasonable to approximate the subject's spine joint locations with the Visible Human Project data because both subjects were similar in size (i.e., less than 4 cm difference in height). A length of 25 mm was then added to the distances obtained from the Visible Human Project to account for the distance from the skin to the marker center. The actual locations of the subject's back joints during the experimental lifts were estimated by translating marker positions in Excel by the appropriate distances and angles.

Additional data processing was also required for the motion capture data. Excel was used to calculate joint angles based on the joint coordinates. A convention was adopted to express angular joint positions and is illustrated in Figure 2.2. The angular positions of each joint were smoothed by fitting the data to a tenth-order polynomial. Hsiang and Ayoub proposed representing angular displacements with eighth-order polynomials [34], but a higher order was used for this study to account for the maximum number of significant local optima exhibited by the subject's joint position trajectories. Angular velocities and accelerations were obtained by differentiating the polynomials with respect to time.

Force platform data was used to determine foot center-of-pressure (COP) throughout the lifts. The COP and ground reaction forces were later compared to those computed by the model in order to validate the model's results. The force platform data was synchronized with the video motion capture data through the use of a button that was pressed immediately before each trial began. The button caused a light to flash that was visible from each camera and also sent a signal to the software recording the force platform data.

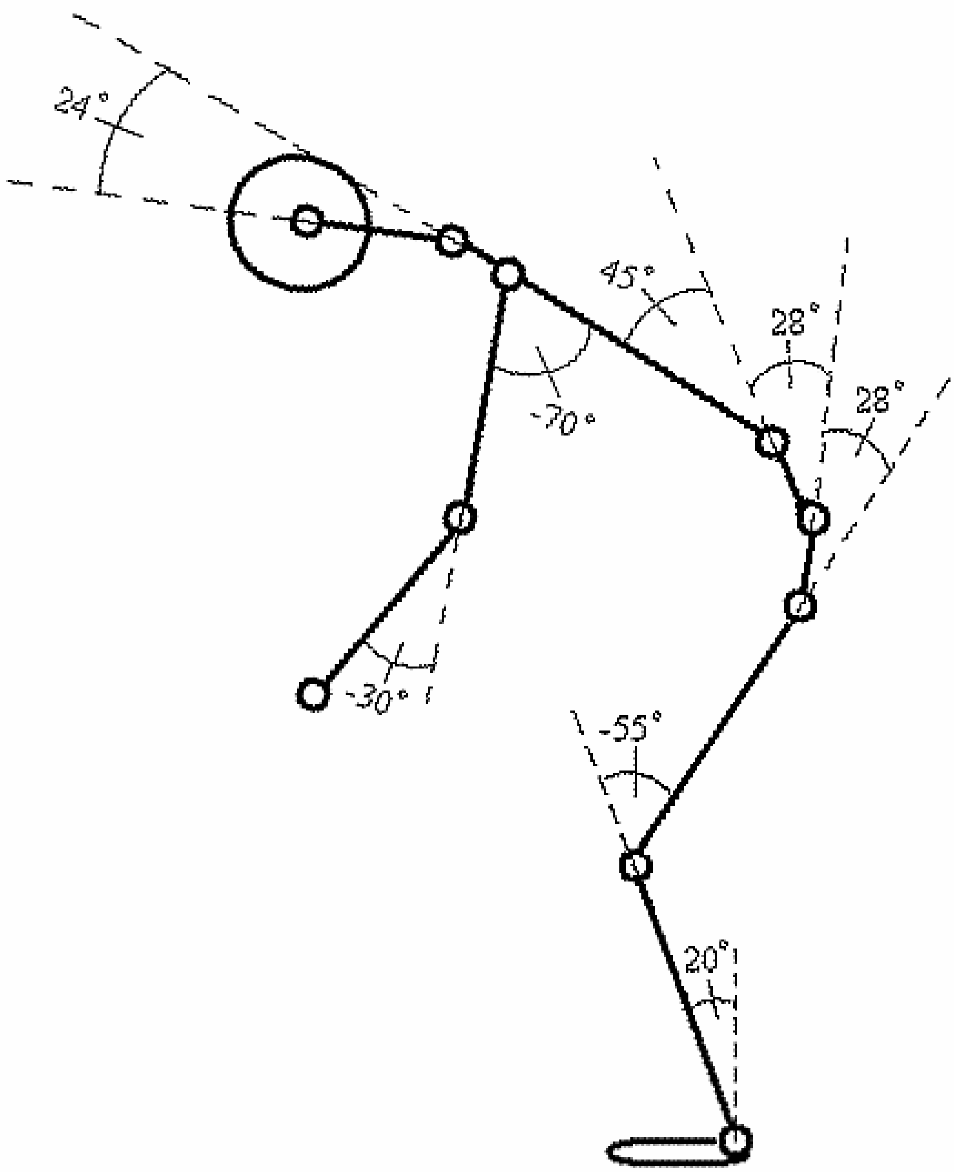


Figure 2.2. Linked model figure demonstrating joints, segments, and angular convention

Great care was taken to define consistent points in the recorded data deemed to signify the start and finish of lift motion for each trial. To identify the beginning point of each trial, the average vertical joint positions were graphed with respect to time. The time in which the average vertical joint positions began to increase indicated the point in which the subject started lifting the weight upward. The end of the lift was established as the time in which force platform data revealed that the COP began to shift heel-ward, signifying that the weight of the load was being supported by the weight rack. Due to the nature of the data, beginning and ending points of the lift were fairly easy to identify by visual inspection. Force platform data was missing for the third medium experimental trial, so the ending point of the lift could not be determined. This trial was therefore omitted from further analysis.

After reducing the data for all eight recorded trials (i.e., three light, three heavy, and two medium), a representative lift was selected for each load, according to the kinematic patterns of the subject as load was increased. Prevailing literature states that people tend to change their lifting technique as they pick up heavier loads. The most commonly documented trends are: (1) for heavier loads, peak lifting velocity tends to be reduced, which causes longer lift durations [12, 17, 18, 29], and (2) for heavier loads, back extension occurs later in the movement than knee extension [12, 15-22]. The selected trials exhibited kinematic trends that most closely reflected those patterns documented in the literature. For each load category, only the trial selected to be most representative of the commonly documented trends was further analyzed and implemented in the simulations.

In order to select representative trials according to the commonly reported velocity and lift duration trends, charts of lifting speed and duration was created to analyze all eight recorded trials (see Appendix A). The lifting speed of each experimental trial was defined as the maximum vertical velocity of the load [15].

Representative trials were also chosen for interjoint coordination patterns by examination of angle-angle plots and angular position time series overplots for the knee and lumbar spine joints. The lumbar spine joint angle was defined as the summed angles of the hip and L4-L5 joints. For both the angle-angle plots and position time series overplots, the knee and lumbar spine angles were normalized from 0 to 1 such that values closer to 1 indicated greater joint extension. Angle-angle plots were helpful in comparing the coordination of joint pairs and also allowed multiple trials to be plotted and compared on the same graph. Position time series overplots offered intuitive visualization of lumbar spine extension lagging behind that of the knee, but separate graphs were required for each of the eight experimental trials (see Appendix B).

CHAPTER THREE

Methods – Model Description

The biomechanical computer model consisted of eight rigid links (i.e., shank, thigh, pelvis, abdomen, thorax, head, upper arm, and forearm/hand/load) that were connected by pin-centered hinge joints (see Figure 3.1). Since lifting naturally occurs almost exclusively in the sagittal plane and the experimental lifting protocol helped to ensure this type of motion from the subject, the model was implemented in two dimensions. The model's ankle was considered to be anchored to the ground since the foot was prohibited from moving in the experimental trials. The forearm, hand, and load were considered as one segment because the slight movement of the wrist was assumed to have negligible impact on the overall lifting dynamics.

Generally, studies have used a minimum of five segments to model planar lifting. The head is often omitted, and the trunk is commonly represented by only one or two segments [5, 13, 32, 34-37]. In these studies, the head is neglected primarily because it is not serially linked between the load being lifted and the ground. However, the head accounts for approximately 8% of total body mass [43]. By modeling the head in the current study, the effects of the head's position on balance were considered. The current study also partitioned the trunk into three segments (i.e., pelvis, abdomen, and thorax) in order to avoid oversimplification of trunk rigidity. In preliminary experiments for the current study, one and two segment trunks were found to be insufficient to describe the back curvature and interjoint coordination demonstrated by subjects, which prevented the model from representing some critical motions of the subjects. Lariviere and Gagnon

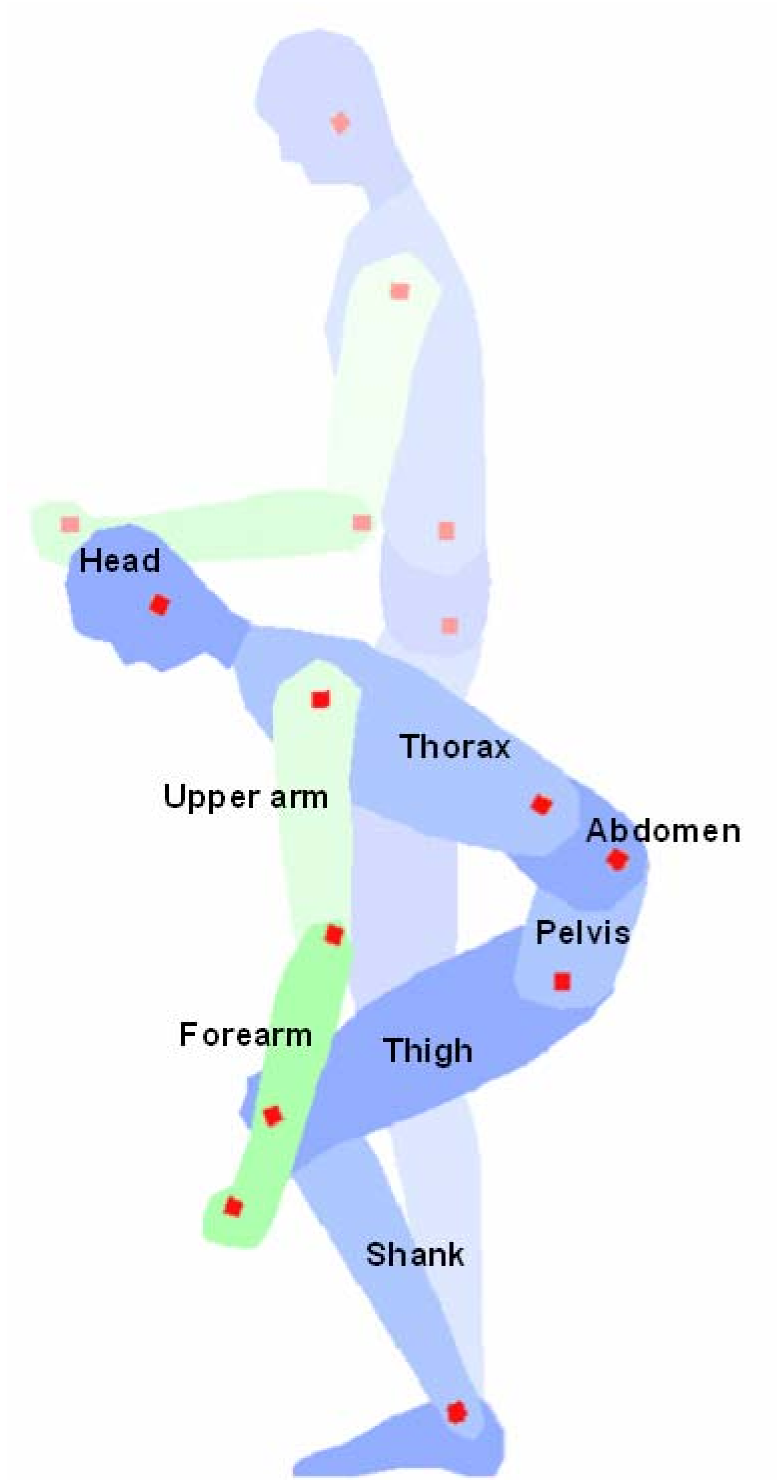


Figure 3.1. Animated planar model illustrating segments and joints in the initial (near) and final (far) postures.

also pointed out that, although the trunk is often modeled as two segments or less for the study of lifting tasks, this assumption must be questioned for any analysis where the trunk is flexed [46].

In order to accurately represent the subject, body segment parameters of the model were assigned corresponding to the subject's segments. Each of the model's links was assigned a length based on the subject's anthropometric measurements shown in Table 2.1. Inertial properties for the model's segments were derived according to the published work of Dempster, who developed proportionality tables to estimate the mass, center of gravitiy, and radius of gyration of body segments for a particular person [43]. These inertial properties were scaled for the subject according to the subject's body mass and measured segment lengths. Radius of gyration was used as an indirect means to calculate mass moment of inertia (I^*) for rotations about the segment's center of gravity [47]. The mass and rotational inertia of the weight were set according to the desired load for a given simulation and were centered about the gripping location of the forearm/hand/load segment. In Table 3.1, the body segment inertial properties that were implemented in the model are provided.

Table 3.1. Inertial properties for each of the model's body segments . Parameters are from Dempster [43] and are scaled to represent the subject being modeled according to his body mass and measured segment lengths.

Segment	Segmental Mass (kg)	Center of Mass (m)	Moment of Inertia (kg m²)
Shank	8.55	0.24	0.27
Thigh	18.38	0.22	0.56
Pelvis	13.05	0.14	0.33
Abdomen	12.77	0.06	0.32
Thorax	19.85	0.07	0.80
Upper arm	5.15	0.12	0.09
Forearm	4.04	0.24	0.21
Head	7.44	0.11	0.02

Since, Dempster did not provide radius of gyration data for the pelvis, abdomen, and thorax, special care was taken to derive the I^* values for these segments. Dempster did provide proportionality data for the radius of gyration for the entire trunk, so this value was used to obtain the scaled I^* for the subject's trunk [46]. The I^* values of each back segment were estimated by distributing the trunk's I^* to the pelvis, abdomen, and thorax based on (1) the distances from each back segment's COM to the COM of the whole trunk and (2) the proportions of the back segments' lengths to the entire trunk length, obtained from the Visible Human Male image set [45].

Dynamic equations of motion were derived based on model parameters using custom software. The software generated computer code in C++ (Metrowerks, Austin, TX, CodeWarrior IDE ver. 5.1.1.1105) that solves equations of motion for the model and can be used for simulating model dynamics.

The model was also constrained according to strength limitation estimates of the subject's joints. Several studies published results defining position-dependent, isometric strength profiles for the primary joints examined in the current study [48-54]. These strengths were all established for "average" males. Since the subject seemed to fit this description, the model was constrained and analyzed according to these same strength limitations. These sources provided agonist and antagonist torque limits for each joint implemented in the model except the C7-T1 joint [48-54]. Lumbar strength data indicated the general strength of the lower back rather than torque capacities of specific vertebral joints [52]. Both the L4-L5 and T12-L1 joints were given this same strength parameter because of their serial linkages together and physiological similarities. The C7-T1 joint was assigned an arbitrarily large (60 Nm) strength parameter in the model because it only needed to support the head and was not serially linked between the load and ground. This seemed justified because no strength limitations were expected to prevent the head from retaining a natural position during the lift.

It is well known that joint strength profiles vary according to position and velocity. Each of the strength sources used by this study contained adequate position-dependent isometric joint strengths, but for many joints, kinetic strength profiles were unavailable. Additionally, low lifting velocities (e.g., relative to ballistic movements such as jumping) did not seem to necessitate isokinetic strength profiling. Therefore the model did not take into account variation in strength due to the joints' angular velocities.

Strength was constrained in the model by using a change of variables technique. The change of variables technique was implemented to accommodate easier interpretation of joint torques, simplify the optimization process, and restrict the model to work within reasonable strength limits. To accomplish the change of variables, torques were represented as control values ranging from 1 (maximum antagonist torque) to 2 (maximum agonist torque). A control value of 1.5 indicated zero torque. Control values ranged linearly from 1 to 1.5 and from 1.5 to 2. Maximum agonist and antagonist torques for each joint were determined from the average male joint strength data obtained from the literature [48-54]. The strengths from these studies were position-dependent, which allowed the control values to also be a function of the joints' angular positions. The change of variables did not allow values below 1 or above 2 and thus, prevented the model from applying joint moments that would exceed the maximum strength for a joint given its angular position.

CHAPTER FOUR

Methods – Simulation of Movement

Once it was shown that the subject exhibited suitable experimental kinematics to be representative of the general lifting population and the model was properly specified to represent the subject, the next step was to use the experimental kinematics as inputs for the model to generate simulations of lifting movement. By properly defining the model and simulation process, simulations were expected to reveal realistic results and behavior pertaining to the physical effects that impact lifting technique. Specifically, the goal of the study was to perform simulations that could isolate potential kinematics that are intrinsically related to load (i.e., independent of intentional influences). While most researchers have presumed that individuals are physically unable to demonstrate identical lifting kinematics if load magnitude is altered [17, 25-28], it may be possible that load does not dictate lifting kinematics, and lifters voluntarily choose to alter their techniques in response to load. To account for this possibility, simulations were evaluated based on energy efficiency, peak joint torque, and stability to determine if some lifting movements adopted by the subject were advantageous over others.

Chapter Four Summary

Typically, computer simulations are implemented through an optimization-based integration process [4, 34]. In the first section, *Simulation Procedure*, the details of how movement is normally simulated through the optimization-based integration process are discussed. Then the specific implementation of the current study's simulation techniques is described. In *Simulated Lifting Scenarios*, the specific situations that were simulated in this study are explained. *Energy Efficiency* contains methods for calculating work and muscular “effort” to evaluate simulation efficiencies. The following section, *Peak Muscular Utilization Ratios*, describes a method of analyzing peak joint torques to assess the impact of joint strength among simulations. In the last section, *Isolating the Effects of Balance on Simulated Lifting*, methods are described that were used to produce another set of simulations that were executed using more stringent requirements in order for the model to maintain stability.

Simulation Procedure

General Simulation Procedure

Simulation is conventionally implemented through a process called integration [4, 34]. First, initial velocity and position states are defined, and the entire movement to be simulated is divided into many brief time intervals. Starting at the first time interval, the model performs forward dynamics analysis, which means that the model solves the equations of motion based on any instantaneous torques or forces that are applied to the model in order to determine the resulting acceleration states. The velocity and acceleration states can then be numerically integrated with respect to time to yield position and velocities for the following time interval. This process is repeated for every time interval to reveal the motion of the model based on the torques and forces that are applied throughout the movement.

The commonly encountered difficulty with the integration process is that the proper instantaneous torques or forces that are needed to simulate the desired movement are often unknown [32, 39]. If the desired movement is already known, then inverse dynamics analysis can be used to estimate the instantaneous forces or torques. For inverse dynamics, the model solves the equations of motion backwards in time to yield the torques or forces necessary to cause the desired accelerations.

Unfortunately, the torques and forces obtained by inverse dynamics analysis are normally insufficient to yield the desired motion when simulated by forward dynamics. The combination of numerical error inherent in the integration process and residual noise from experimental data collection tends to compound and causes dramatic impacts on simulation results. Typically, researchers have attempted to resolve this problem by tweaking the inverse dynamics torques to be suitable for forward dynamics simulation by using optimization algorithms to select torques according to performance criteria that are presumed to represent an optimal strategy [5, 32, 34, 36, 37, 39].

Inverse Dynamics Analysis

Once the model was fully specified, inverse dynamics analysis was applied to estimate the joint torques necessary to produce the motion of the subject's experimental trials. As mentioned previously, the experimental kinematics were available in the form of tenth-order polynomials representing angular positions, angular velocities, and angular accelerations for each joint. In order to approximate instantaneous torques throughout the lifting movement, the experimental trial was partitioned into time intervals of 1/60 seconds. These time intervals were referred to as control steps. Since 60 Hz video cameras were used for the motion capture, the duration of each experimental trial was evenly divisible by the number of control steps, which eliminated difficulties in boundary conditions for the control steps. At each control step, the polynomials were evaluated at that point in time, and each of the model's joints were set to the appropriate angular positions and velocities. Next, the model solved the equations of motion (i.e., inverse dynamics) to determine the necessary joint torques that would be required to achieve the angular accelerations of each joint according to the acceleration polynomials. The change of variables technique discussed in Chapter 3 was used to convert the joint torques to control values between 1 and 2. This process was repeated for each control step to provide an estimate of instantaneous control values needed to match the experimental kinematics of the subject.

As a check to ensure that the model's equations of motion were solved correctly, time-independent forward dynamics analysis was applied to the results of the inverse dynamics. For each control step, the model's joint positions and velocities were assigned to correspond to the position and velocity polynomials evaluated at that point in time. Then the control values obtained from the previous inverse dynamics analysis were converted back into the torques and were applied to the joints. The equations of motion (i.e., forward dynamics) were solved to determine the resulting accelerations. The ability of the model to correctly solve the equations of motion was validated when the accelerations obtained via time-independent forward dynamics were equivalent to those of the acceleration polynomials evaluated at the given point in time.

Forward Integration

Simulations of movement were implemented via numerical integration of the equations of motion forward in time (forward dynamics). As explained above, the lift duration was divided into control steps lasting 1/60 seconds. Because numerical integration yields more accurate results for smaller time intervals than the control steps, each control step was further divided into 20 integration steps. For the first integration step, the initial angular positions were specified to correspond to the subjects initial posture, and velocities of each joint were set to zero because the lifter was considered to be stationary at the start of the lift. Control values corresponding to the first control step were then converted into torques via the change of variables. Forward dynamics used the current position, velocity, and torque states to solve for the resulting joint accelerations. A four-step Runge-Kutta method was then used to numerically integrate the resulting accelerations and velocities in order to calculate the states of the positions and velocities for the next integration step. Since control values were only available for every control step (i.e., as opposed to every integration step), control values to be used by forward dynamics between control steps were obtained by linearly interpolating between the control values of the previous and subsequent control steps. This process was repeated for every integration step to simulate the lifting movement throughout the duration of the lift.

Optimization

In order to obtain appropriate control values to allow successful forward dynamics simulation of the desired motion, a particle swarm optimization algorithm [41, 42] was utilized to refine the initial torque estimates that were determined by inverse dynamics analysis from the experimental kinematics. The particle swarm algorithm used 20 “agents” to search the space of all combinations of control values to identify an “optimal” solution, which best tracked experimental kinematics. Each agent generated pseudo-random control values between 1 and 2 for each of the model’s joints. By repeatedly calling the objective function for different sets of control values, the algorithm was able to hone in on the "optimal" solution.

Rather than attempting to optimize all the control values needed for the entire forward integration process at once, the optimization routine was called repeatedly, and independently, for each control step throughout the lifting simulation. That is, the control values at one control step were optimized to minimize the positional kinematics error simulated within the duration of the two control steps influenced by those control values. For example, the control values of control step 5 were optimized to minimize error accumulated during simulation of control steps 4 and 5. Both control steps are influenced by step 5 control values due to interpolation between control steps. Subsequently, control values at control step 5 were assumed fixed, and the optimization process proceeded on to optimize control values at control step 6. This technique of performing a complete and independent optimization at every control step yielded superior results than optimizing several control steps at once because the integration process is inherently dependent on previous kinematic states.

The objective function used by the particle swarm algorithm to obtain "optimal" control values was chosen to minimize the summed-square positional error accumulated over each control step. By optimizing one control step at a time, the net result was that the summed-square positional error accumulated over the entire lift duration was minimized. The objective function is given in Equation 4.1 below.

$$\text{Minimize} \quad \int_0^T \sum_{j=1}^{8} \left[a \cdot X_j(t)^2 + a \cdot Y_j(t)^2 + b \cdot M_j(t)^2 \right] + 4 \cdot a \cdot W(t)^2 \quad dt \tag{4.1}$$

where:

T is the total time duration of the lift,

j is the identification index for each joint,

$X_j(t)$ is the joint positional error in the horizontal direction at time t,

$Y_j(t)$ is the joint positional error in the vertical direction at time t,

$W(t)$ is the sum of the magnitudes of the horizontal and vertical positional errors of the weight at time t,

$M_j(t)$ is the joint moment error at time t,

a and b are error weighting coefficients.

The positional error refers to the difference between the joint coordinates of the actual lift performed by the subject and those of the simulated lift. The moment error refers to the difference between the simulated torques (obtained from the optimized control values) and the torques required to yield the experimental kinematics (calculated by inverse dynamics analysis). Moment error was incorporated in the objective function because the optimization routine consistently provided lower summed-square positional error with the addition of the moment error than without it. This effect was observed because the moment error minimized oscillation of the joint torques, which prevented high velocities that tended to overshoot the desired positions. Weighting coefficient values were obtained by trial and error to determine the smallest summed-square positional error results. Coefficient values for “a” and “b” were 1,000,000 and 300, respectively. Therefore, moment error was substantially less significant than positional error.

Simulated Lifting Scenarios

Once the computer model and optimization-based simulation techniques were properly implemented, simulations were performed in which the model was assigned a particular virtual load and tasked with reproducing the kinematics of the subject lifting his actual load. A matrix of nine simulations resulted from the different model weights and experimental data sets. For three scenarios, the model's virtual load was the same as the actual load. These scenarios simply modeled the subject exactly as he performed actual lifts. For the remaining six scenarios, the model attempted to match kinematics obtained from the subject lifting an actual load that was different than the model's virtual load (e.g., the model lifts the heavy weight while trying to follow the kinematics of the subject lifting the light weight). These simulated scenarios allowed direct comparison of the kinetics and kinematics generated by the model so that the question, "Does lifting motion change voluntarily or intrinsically as load is increased?" might be answered.

A naming convention was adopted to quickly reference individual lifting scenarios in the matrix. The scenarios were named according to their virtual (load specified to model) and actual (load lifted by subject for target kinematics) loads. The virtual load was denoted by a "v" and was followed by "L", "M", or "H", corresponding to whether the virtual load was light, medium, or heavy. Similarly, the actual load was denoted "a" and was also followed by "L", "M", or "H" (e.g., the scenario in which the model lifted the light virtual load while trying to follow the kinematics of the subject lifting the heavy actual load is named vLaH). This naming convention is shown in Figure 4.1 for all nine simulations.

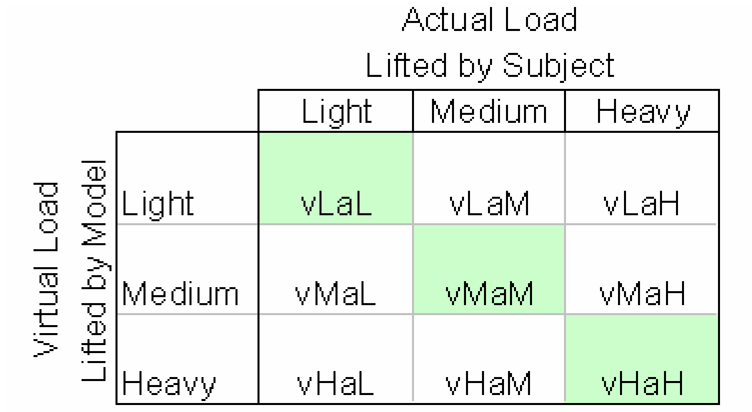


Figure 4.1. Matrix of simulated scenarios performed by the model. For each scenario the model lifted a light (vL), medium (vM), or heavy (vH) virtual load, while tracking experimental kinematics of the subject lifting a light (aL), medium (aM), or heavy (aH) actual load. Diagonal elements (shown in green) indicate scenarios in which the virtual and actual loads were identical.

Energy Efficiency

Measures of energy efficiency were computed for each matrix scenario to determine if some lifting movements were more advantageous than others. Muscle work is commonly computed in biomechanical simulations because the elements needed are readily attainable by the model. Work can be obtained by integrating power, and power can be calculated for a given joint simply by taking the product of the joint's moment and angular velocity. The calculation of muscle work for all the joints is defined according to Equation 4.2. The total muscle work performed on every joint is also equivalent to the mechanical work done on the total COM (i.e., combined COM of all the body segments and load), which can be calculated in another way. This calculation of work is performed by multiplying the vertical displacement of the total COM by the gravitational force acting on the total COM as shown in Equation 4.3. While these two computations of work are equivalent, they offer two different perspectives concerning the way work is performed on the system.

Muscle Work = $\int_0^T \sum_{j=1}^{8} \left[M_j(t) \cdot v_j(t) \right] \, dt$ (4.2)

where:

T is the total time duration of the lift,

j is the identification index for each joint,

$M_j(t)$ is the joint moment at time t,

$v_j(t)$ is the joint velocity at time t.

Mechanical Work = $m \cdot g \cdot (y_f - y_i)$ (4.3)

where:

m is the mass of the total COM,

g is the gravitational acceleration constant,

y_f is the final vertical position of the total COM,

y_i is the initial vertical position of the total COM.

Muscular “effort” was also estimated to provide insight into the energetic efficiency of each lift. Effort is a somewhat subjective term that is often used to describe the level of undertaking a person exerts during a task. “Effort” was quantified according to the sum of squared ratios of joint moment and strength as is given in Equation 4.4. In several lifting studies, researchers have implemented this calculation as the objective function for optimizing simulations [5, 32, 34, 36].

“Effort” = $\int_0^T \sum_{j=1}^{8} \left[\frac{M_j(t)}{S_j(\theta_t)} \right]^2 \, dt$ (4.4)

where:

T is the total time duration of the lift, j is the identification index for each joint,

$M_j(t)$ is the joint moment at time t,

$S_j(\theta_t)$ is the joint moment strength for a position θ at time t.

Equation 4.4 expressed muscular “effort” in terms of the ratio of instantaneous torque to maximum joint strength. Essentially, each joint moment at a given point in time was normalized by the position-dependent strength of the joint at that time. The joint strengths were based on average male strength data obtained from various sources in the literature [48-54] as is discussed above. Gagnon and Smyth first used this measure and coined the term, muscular utilization ratio (MUR), to express how severely the joint was “being needed” throughout the movement [55]. Equation 4.4 quantifies the “effort” required for a lift based on the time duration of the lift and the joint moments relative to their potential capacities.

Peak Muscular Utilization Ratios

Peak MUR represented the ratio of peak joint torque to joint strength and was intuitively expressed as a percentage. For each matrix scenario, peak MUR was examined for each joint to identify which joints underwent the greatest torques relative to their strengths. Additionally, the magnitudes of the MUR’s indicated the extent to which a joint’s maximum torque approached the joint’s strength capability. Furthermore, the MURs of alternate scenarios were compared to observe how peak torques were affected in response to the model lifting different virtual loads and generating different experimental kinematics.

Isolating the Effects of Balance on Simulated Lifting

As the computer model simulated lifting movement for each scenario, foot COP was also computed and monitored to ensure that appropriate stability was constantly maintained. The model's COP was calculated according to the ground reaction forces derived from gravitational and inertial accelerations of the total COM of the lifter and load. Therefore, COP represented the horizontal position at which a resultant ground reaction force would have to be applied to the foot in order to keep the model in equilibrium (i.e., horizontal location of ground reaction force to prevent model from tipping over). Experimental force platform data was used to validate the COP computed by the model for situations in which the virtual and actual loads were the same. It was assumed that adequate stability would be maintained as long as the COP remained within the base-of-support (BOS), as demonstrated by Equation 4.5 [5, 33, 36, 37]. The length of the BOS corresponded to the distance from the subject's heel to his toe (see Table 2.1), in accordance with the methods from other simulated lifting studies [33, 36, 37]. By monitoring the position of the COP during simulation, it was determined that for all nine matrix scenarios the conditions of balance were never violated. Therefore, each matrix scenario demonstrated stability throughout the movement without the addition of balance constraints on the model.

$$X_{toe} \leq COP \leq X_{heel} \tag{4.5}$$

where:

X_{toe} is the horizontal position of the end of the toe,

COP is the position of the foot center of pressure,

X_{heel} is the horizontal position of the end of the heel.

To further test the model's ability to preserve stability, another matrix of nine lifting scenarios was generated. This set of scenarios is referred to as the "balance matrix". During the generation of the balance matrix, the BOS was reduced and a penalty was implemented in the objective function to prevent the COP from moving beyond the BOS. The penalty increased exponentially as the COP approached the BOS limits such that the optimization process enforced stability. If the COP was not near the BOS limits, the balance penalty was not added to the objective function, and the optimization process continued as it did for the original matrix scenarios.

Special care was taken in the choosing of the reduced BOS parameter. The reduced BOS was chosen to highlight potential kinematic differences required to maintain stability in balance matrix scenarios. Since the subject was able to maintain balance during the experimental trials, the reduced BOS also had to be chosen such that the diagonal balance matrix scenarios (i.e., with identical virtual and actual loads) would be able to simulate movement without incurring the balance penalty. Therefore, the reduced BOS was chosen by shortening the distance from the ankle to the toe as much as possible while still allowing the balance matrix vHaH scenario to simulate without incurring the balance penalty. The reduced BOS length from the ankle to the toe was 0.17 m compared to the subject's ankle to toe length of 0.21 m. Every scenario in the balance matrix used this same reduced BOS parameter. The reduced BOS value was based on simulation results of the vHaH scenarios because: (1) the BOS parameter needed to be specified according to a realistic lift (i.e., identical virtual and actual loads) as opposed to a hypothetical scenario, and (2) the COP was expected to be closest to the toe for the heavy load since the magnitude of the heavy load would naturally shift the horizontal component of the total COM toward the toe. For this reason, vH scenarios were expected to be more sensitive to stability problems than scenarios where lighter loads were lifted.

CHAPTER FIVE

Results

The purpose of the current study was to examine the dynamics of lifting through the use of a biomechanical computer model. Lifting motion data from experimental trials of a human subject was used as an input for the model. Body segment parameters corresponding to the subject were specified to the model so that it could accurately represent him. It was anticipated that the model might determine whether or not the documented kinematic patterns that tend to consistently change in response to load can be isolated. In addition, the model was used to analyze factors that may cause lifting kinematics to change as a function of load.

Chapter Five Summary

Through the course of the study, important findings were obtained in several areas of investigation. Considerable published data has revealed that subjects tend to adjust their lifting strategies in the same way as the magnitude of the load being lifted is altered [12, 15-22]. Since this assumption laid the foundation for the current study, detailed analysis was undertaken to extract similar kinematic trends from the experimental trials of the subject. In the *Experimental Kinematics* section, the level to which the subject demonstrated the expected patterns is described. The ability of the model to track the experimental kinematics is discussed in *Comparison between Simulated and Experimental Kinematics*. These findings address the primary motivation for the study. The proficiency of the model to calculate similar ground reaction forces as were obtained by force platform data is given in *Comparison between Simulated and Experimental Ground Reaction Forces.* In the next section, *Model Stability*, the model's capacity to match the experimental kinematics under conditions requiring superior stability is presented. The effect of balance on lifting technique is highlighted by analysis of these results. In *Joint Torque Analysis,* data is provided for all nine simulated scenarios indicating the extent to which each joint underwent maximum exertion. The last section, *Energy Efficiency*, contains the results of muscle work and "effort" calculations for each simulated scenario to indicate relative advantages in efficiency among the scenarios.

Experimental Kinematics

As described in Chapter 2, three experimental trials (i.e., one light, one medium, and one heavy) were chosen to be representative of the general lifting population. The selected trials were chosen to best reflect the trends most commonly reported in the literature. The experimental motion-capture data revealed that, for the selected trials, the subject followed coordination patterns similar to those that are commonly documented. Namely, these trends are: (1) as heavier loads are lifted, lifting velocity is reduced, which causes lift duration to increase [12, 17, 18, 29], and (2) back extension occurs later in the movement compared to knee extension as the load increases [12, 15-22]. While these coordination patterns were not reflected for every combination of the light, medium, and heavy experimental trial, the selected trials exhibited excellent agreement between the subject's motion and the anticipated tendencies.

Desirable patterns were recognized for both lift velocity and duration for the selected trials. The reduction in peak vertical velocity of the load as heavier weights were lifted can be identified in Figure 5.1. A rise in lift duration corresponding to increasing load magnitude is evident from Figure 5.2.

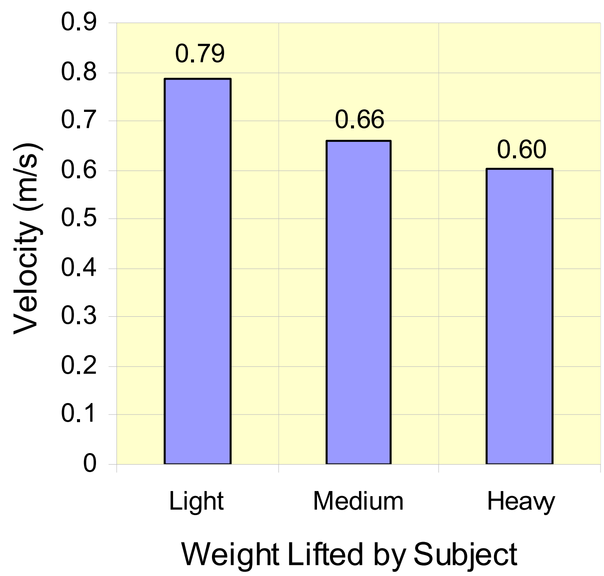


Figure 5.1. Peak velocity for each representative trial performed by the subject. These trials reflect a clear trend that as load is increased, peak velocity is reduced. This pattern is supported by the prevailing literature [12, 17, 18, 29].

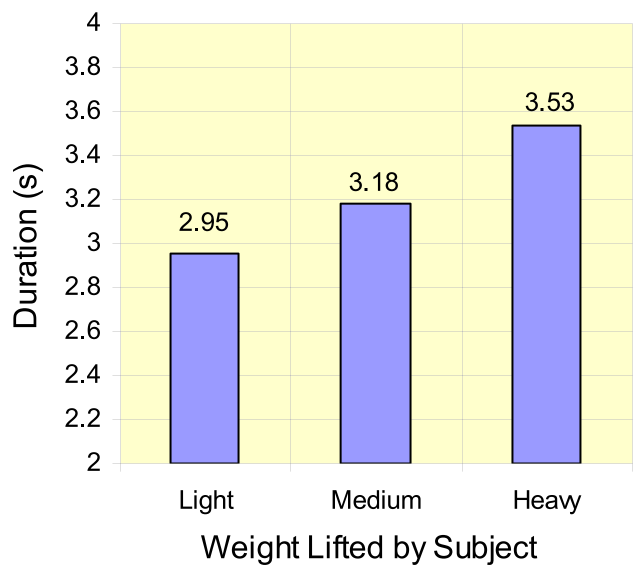


Figure 5.2. Lift duration for each representative trial performed by the subject. These trials reflect a clear trend that as load is increased, lift duration increases also. This pattern is supported by the prevailing literature [12, 15-22].

Agreeable interjoint coordination patterns between the knee and back were also observed by angle-angle plots. The interjoint coordination relationships between the lumbar spine and knee joints adapted from prevailing literature [17] can be compared to those of the subject in Figure 5.3. The amplitudes of the angular positions were normalized from 0 to 1 for both joints, denoting that higher values indicated greater joint extension. For both angle-angle plots, it is evident that the back extends increasingly later than the knee as heavier loads are lifted. This trend is evident because, starting at the bottom left corner of the curve, the knee angle increases (indicating extension) before the lumbar spine angle increases, and this pattern is exaggerated for heavier loads.

The published knee-lumbar relationship shown in Figure 5.3 (A) seems to have a more pronounced pattern as load increases than the results of the current study (Figure 5.3 (B)). This may be due to the use of squat technique in that study [17]. In the squat lift, the subject is forced to lift initially with only knee extension. In the initial posture, the back is straight but tilted forward slightly to prevent the load being lifted from colliding with the knees. Once the knees are nearly straight, the squat technique requires that the back extend from the initial angle of incline to finish the movement in the erect posture. Therefore, the squat technique naturally causes the exaggeration of lag time between the knee and lumbar. The current study still displayed the desired knee-lumbar relationship in response to load, but the freedom of the subject's self-chosen freestyle technique did not necessitate a delay before the back could extend completely.

The desired patterns of interjoint coordination are also demonstrated in Figure 5.4 via position time series overplots of the knee and lumbar spine joints. These overplots are useful to describe when joints extend relative to each other in a given trial, but

A.

B.

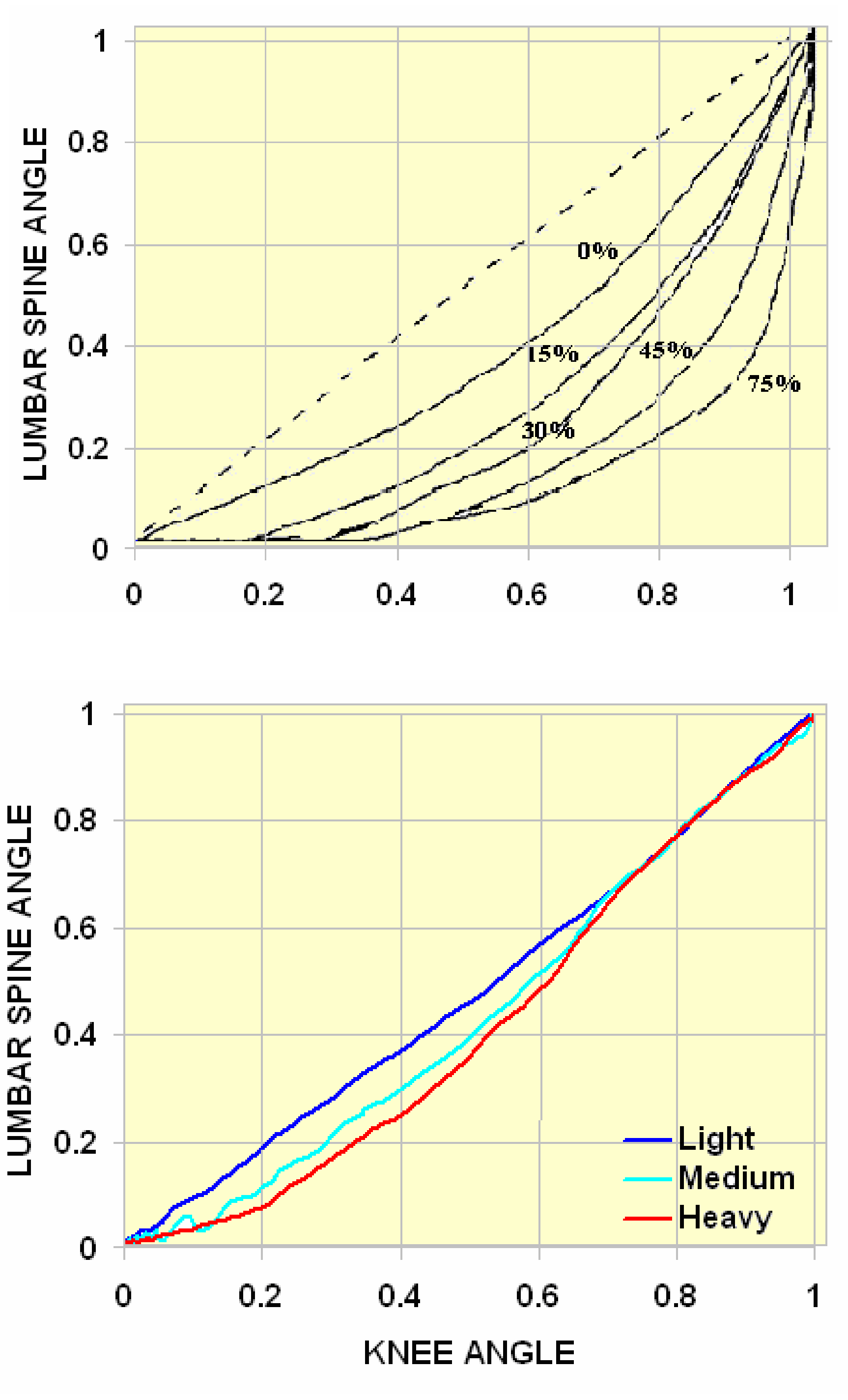
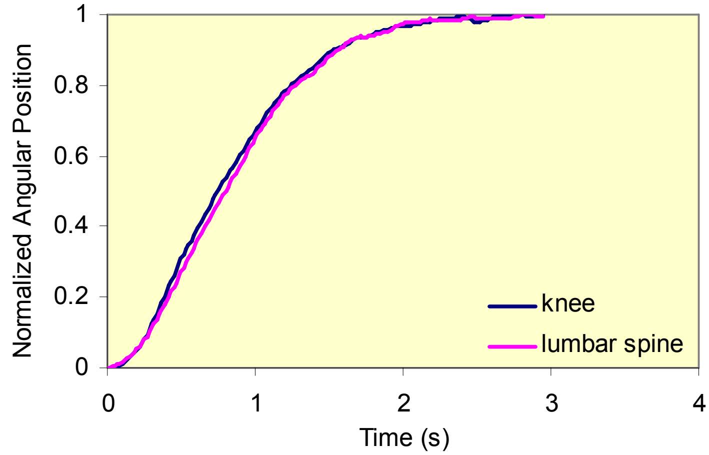


Figure 5.3. Comparison of interjoint coordination between (A) data adapted from Scholz [17] and (B) experimental data from the current study's subject. Interjoint relationships are expressed as normalized angle-angle plots of lumbar versus knee. The percentages in (A) denote the loads lifted relative to MAWL. Both plots exhibit the documented trend of back extension lagging increasingly behind knee extension as heavier weights are lifted. The more pronounced trend in (A) is most likely due to the squat technique being prescribed.

A.



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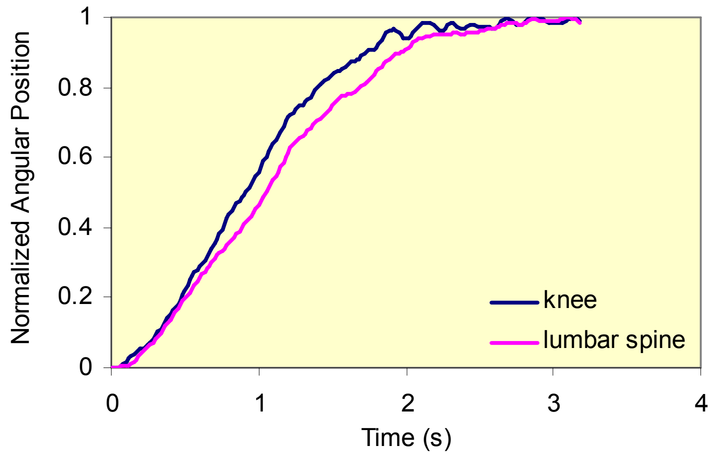


C.

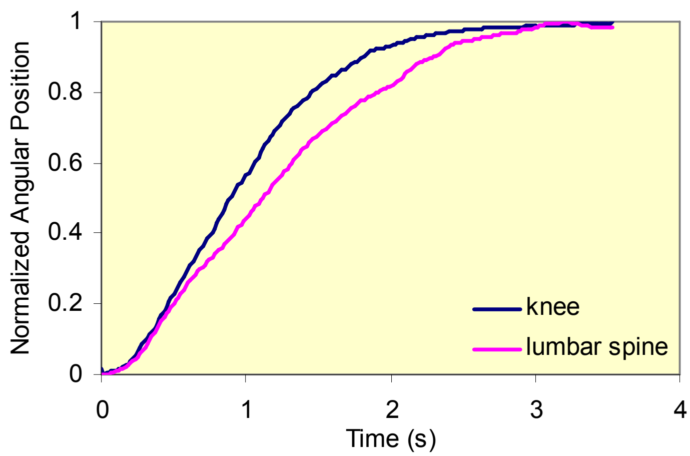


Figure 5.4. Normalized angular position time series overplots of the knee and lumbar spine joints for representative experimental lifting kinematics of the (A) light, (B) medium, and (C) heavy loads. The plots exhibit the documented trend of back extension lagging increasingly behind knee extension as heavier weights are lifted.

multiple overplots must be compared to understand the impact of load on joint coordination. The angular positions of the knee and lumbar spine are also normalized from 0 to 1 such that values closer to 1 indicate greater joint extension. From Figure 5.4 it is apparent that as time increases during the lift, both the knee and lumbar spine extend, and for heavier loads, the knee extends sooner.

Comparison between Simulated and Experimental Kinematics

Contrary to previous popular belief [17, 25-29], the current study's model indicated that the magnitude of the load does *not* dictate lifting kinematics. Despite constraints due to the laws of physics, joint strength, and stability, all nine simulated scenarios were able to track the subject's experimental kinematics with essentially zero deviation.

Figure 5.5 contains time series plots of experimental and simulated angular kinematics in terms of position and velocity for the elbow joint from the vMaH scenario. Similar plots for the remaining joints from the vMaH scenario are provided in Appendix C. The vMaH scenario was chosen to illustrate the ability of the model to track experimental kinematics under conditions of different virtual and actual loads. Other scenarios demonstrated similar results in terms of kinematic deviation. These graphs allow direct visualization of the model's ability to track the experimental kinematics with essentially zero deviation.

Figure 5.5 also contains a time series plot of joint torque so that the experimental torques determined by inverse dynamics can be compared to the simulated torques obtained from optimized forward dynamics analysis. Maximum position-dependent isometric strength under conditions of flexion and extension are illustrated to be

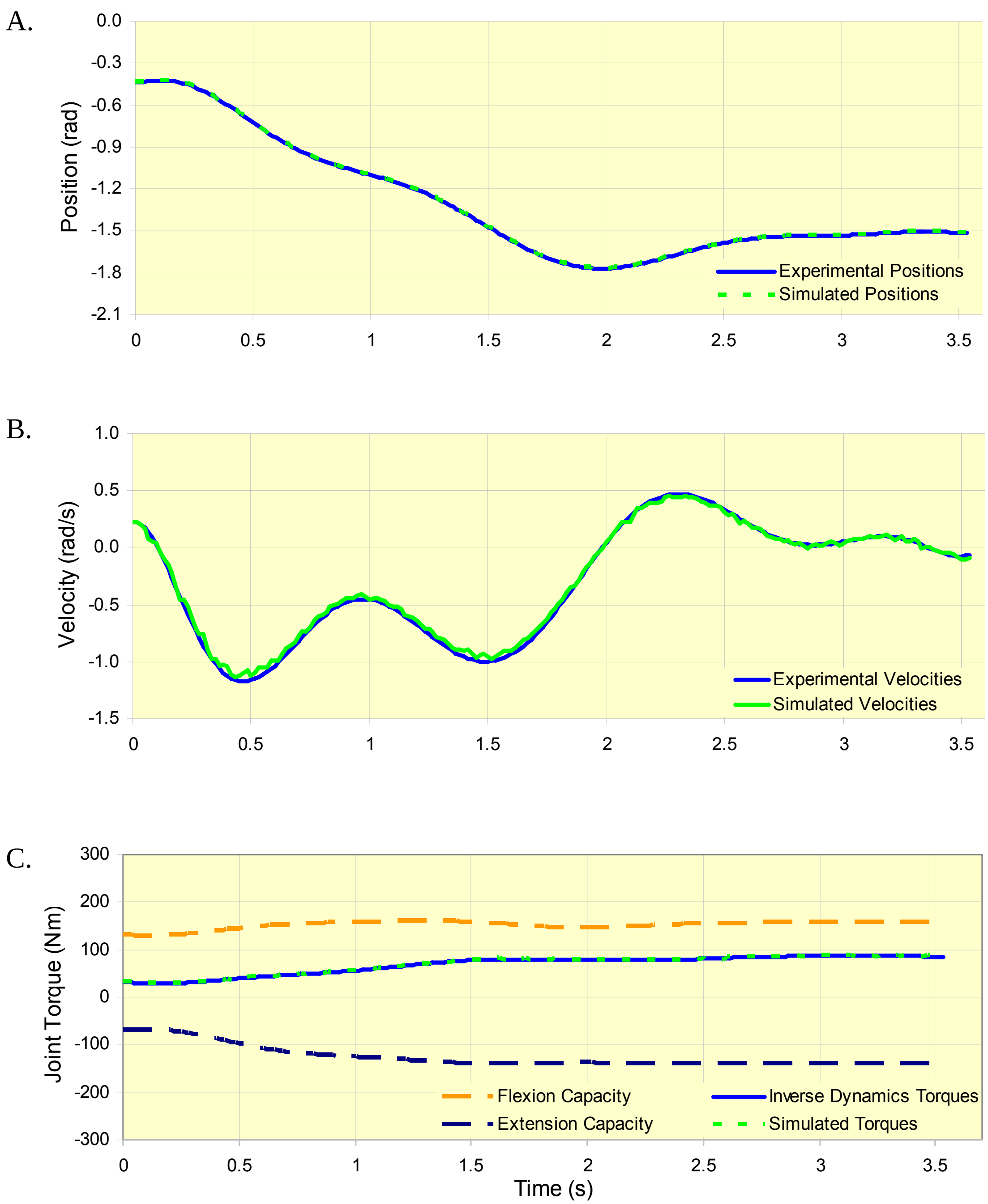


Figure 5.5. Kinematic and kinetic data for elbow joint of simulation vMaH. (A) Angular position time series plot. (B) Angular velocity time series plot. (C) Bound torque time series plot. Bound torque reflects the change of variables discussed in Chapter 2. The plots illustrate minimal deviation between the experimental and simulated data.

compared with the instantaneous torques. One may note that the simulated and experimental moments also exhibit minimal deviation from each other. Appendix C also contains torque deviations for the remaining joints in the vMaH scenario.

A single numerical result was also calculated for each scenario to evaluate the ability of the model to track the subject's experimental kinematics. Throughout the lift duration, the simulated and experimental joint coordinate positions were calculated, and the distances between the positions of the simulated and experimental joints were determined. Differences between the simulated and experimental kinematics of each scenario were quantified as the maximum positional deviation in millimeters among all joints. As Figure 5.6 reveals, among all the joints in all the scenarios, the maximum deviation from the experimental kinematics was only 1.11 mm. Furthermore, this deviation may be neglected due to the nature of the optimization algorithm as explained below.

Nominal deviations in kinematics were anticipated due to the inherent nature of optimization routines to provide a solution that can only closely approximate the theoretical optimum [41, 42]. The deviations expressed in Figure 5.6 were attributed to the optimization algorithm and were not due to any inherent limitations of physics, strength, etc. In support of this assertion, note that the diagonal matrix scenarios, which simulated the subject exactly as he performed actual lifts, exhibited deviations comparable to other scenarios. For all nine scenarios, the maximum positional deviations out of all the joints were also negligibly small (approximately 1 mm). Since the model was expected to generate the subject's realistic movement for the diagonal scenarios, these results helped to validate the model and optimization-based simulation procedure.

Virtual Load Lifted by Model \ Actual Load Lifted by Subject	Light	Medium	Heavy
Light	1.03	0.63	0.91
Medium	0.81	0.71	1.11
Heavy	1.11	1.04	0.68

Figure 5.6. Maximum deviation in millimeters out of all the joints computed for each matrix scenario. Diagonal scenarios demonstrate that deviation caused by optimization limitations is on the order of 1 mm. None of the cases incurred relevant digression from the experimental positions.

Comparison between Simulated and Experimental Ground Reaction Forces

The validity of the model was also supported by strong agreement between the model's computed ground reaction forces and those measured by the force platform data for the three matrix-diagonal scenarios (i.e., same virtual and actual loads). Figure 5.7 contains foot COP and vertical ground reaction force data computed by the model and measured by the force platform for the vMaM scenario. Similar figures for the vLaL and vHaH scenarios are provided in Appendix D.

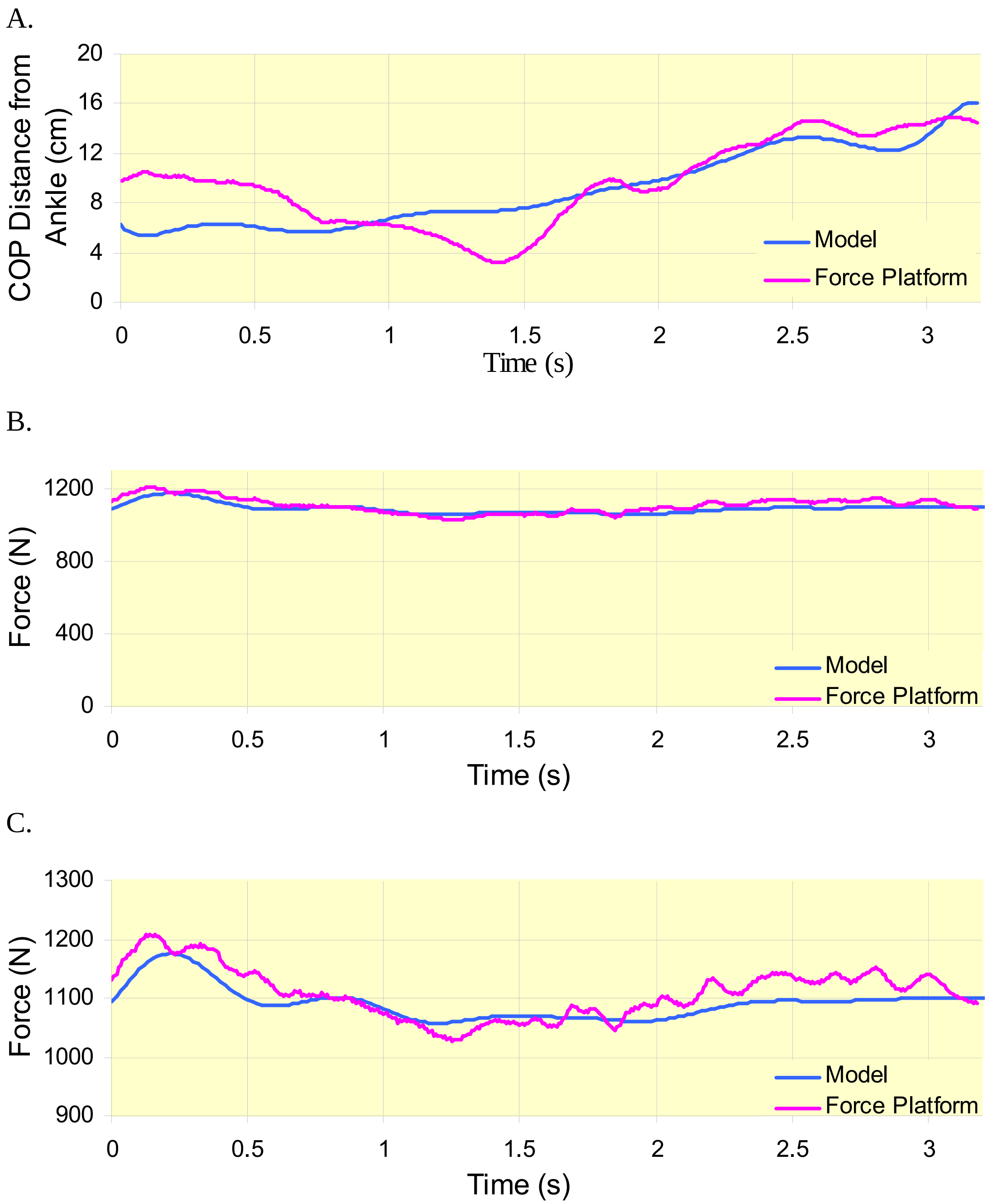


Figure 5.7. Force platform and model trajectories of (A) COP relative to ankle and (B) vertical ground reaction force and (C) vertical ground reaction force scaled up to range 1000 N to 1300 N for scenario vMaM. Positive distances from the ankle indicate that the COP is toward the toe. These trajectories indicate high correlation between the experimental data measured by the force platform and the simulated data derived by the model, which supports the model's validity.

Model Stability

By monitoring the position of the foot COP, it was determined that none of the original matrix scenarios violated the conditions required to maintain stability. In fact, force platform data measured from the subject agreed with the COP computed by the model that for the experimental trials the subject's COP never exceeded 80% of the distance from the ankle to the toe (i.e., the balls of the feet). This finding can be observed from foot COP trajectories in Appendix D. As described in Chapter 4, another set of nine scenarios, called the balance matrix, was simulated to further examine the effects of stability on lifting kinematics. These scenarios were constrained to maintain COP within a reduced BOS. Figure 5.8 contains the trajectories of the COP computed by the model for the original and balance matrices. Analysis of Figure 5.8 indicates that, for the original matrix, only the vHaL and vHaM scenarios had COP trajectories that exceeded the reduced BOS, but all scenarios remained within the original BOS.

The resulting kinematics of the balance matrix revealed that only the vHaL and vHaM balance matrix scenarios incurred the balance constraint penalty, which forced the model to compensate by shifting the body's COM heel-ward near the end of the lift. The remaining scenarios were able to maintain stability within the reduced BOS without incurring significant deviation from the experimental kinematics. Figure 5.9 depicts two overlapping clips of the computer model animation. The back (red) figure represents the final position of the model in the original matrix vHaM scenario, and the front (blue) figure indicates the final position of the same scenario in the balance matrix. From Figure 5.9, it appears that the balance matrix vHaM scenario was forced to compromise "optimal" experimental kinematics in order to maintain stability. Deviation of the

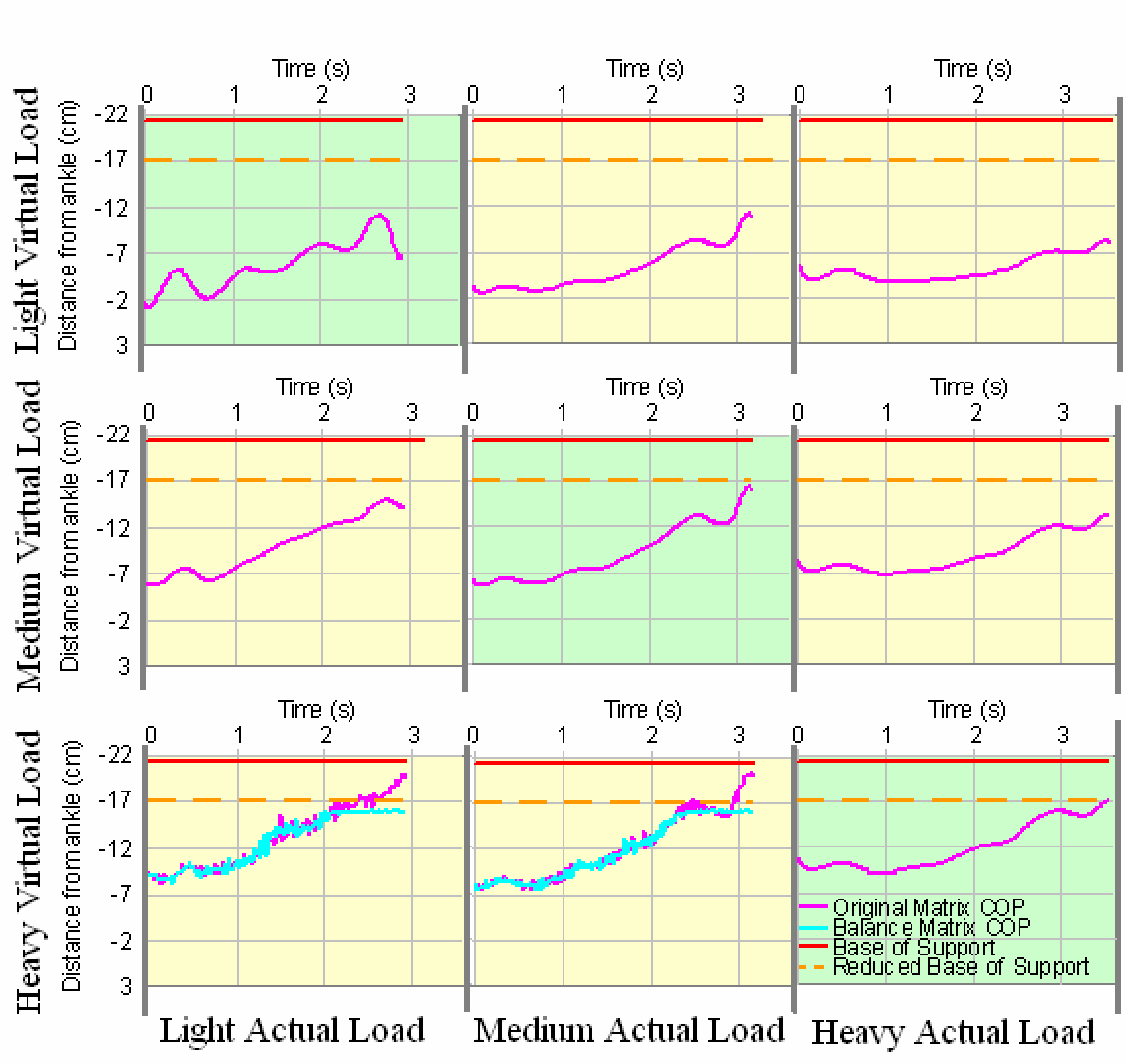


Figure 5.8. Comparison of COP trajectory computed by model to original BOS and reduced BOS for each original matrix scenario. Negative distance from ankle indicates direction is toward the toe. Every scenario's COP remains within the original BOS, but the vHaL and vHaM scenarios exceed the reduced BOS near the end of the lift.

simulated balance scenarios from the experimental kinematics was quantitatively assessed by calculating the maximum positional deviation in millimeters out of all joints for each scenario. Results from this analysis are provided in Figure 5.10. As discussed in the *Kinematic Differences* section, approximately 1 mm of deviation was attributed to limitations of the optimization process. Therefore, analysis of Figure 5.10 suggests negligible deviation in all balance matrix scenarios except for vHaL and vHaM. These scenarios experienced maximum deviations of a few centimeters. These deviations were small, yet significant.

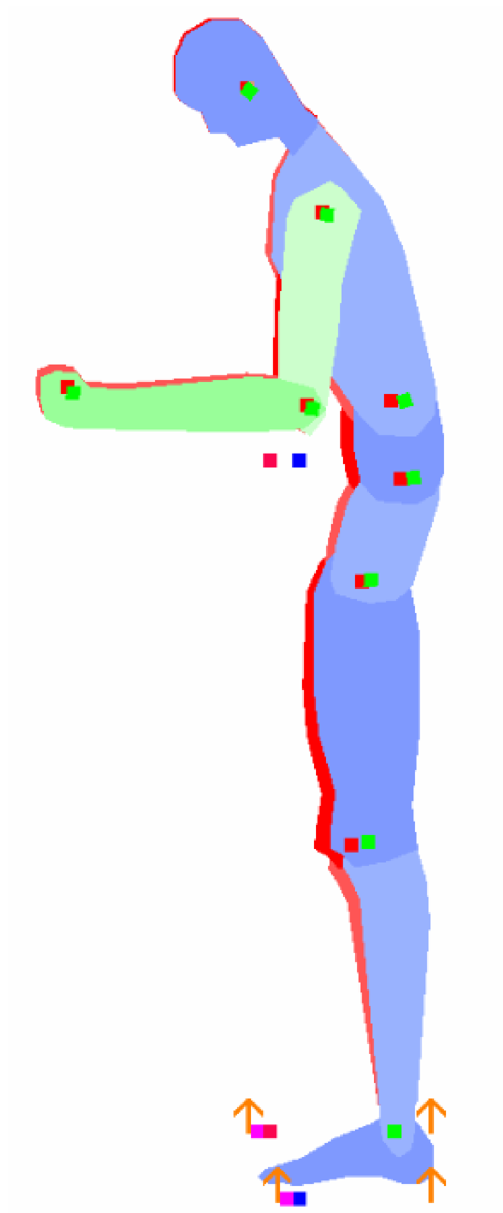


Figure 5.9. Animated simulation comparison of vHaM scenario at the end of the lift for balance matrix (Blue, near) and original matrix (Red, far). Green joints correspond to the balance vHaM scenario. Arrows denote original BOS (Top) and reduced BOS (Bottom) regions. COM positions are marked by blue (Balance Matrix) and red (Original Matrix) boxes. Pink boxes indicate COP. Animation indicates balance matrix simulation shifted body's COM heel-ward.

Virtual Load Lifted by Model	Actual Load Lifted by Subject: Light	Medium	Heavy
Light	0.94	0.90	0.87
Medium	0.72	0.73	0.88
Heavy	36.1	23.7	2.83

Figure 5.10. Quantitative assessment for each scenario in balance matrix of maximum deviation in millimeters out of all joints from experimental kinematics. Only vHaL and vHaM incurred relevant deviation from experimental positions.

Joint Torque Analysis

Joint torques were examined for each scenario in the original matrix according to peak MUR level. Peak MUR levels express the percentage of peak joint torque experienced during the simulation relative to position-dependent joint strength. Figure 5.11 provides each joint’s torque peak MUR level for all nine lifting scenarios.

Several patterns were expected to be revealed by Figure 5.11. With the exception of the C7-T1 joint, peak MUR levels consistently increased in response to the model lifting heavier virtual loads. Torques from the C7-T1 joint was not anticipated to increase with load because it only served to support the head. C7-T1 MUR levels were also low relative to other joints because this joint was assigned an arbitrarily large strength value, as is discussed in Chapter 2.

Figure 5.11 also contains some less obvious items of interest. For a given virtual load, the model generally achieved similar peak MUR levels as it tried to track the kinematics of the subject lifting the light, medium, and heavy actual loads. The medium

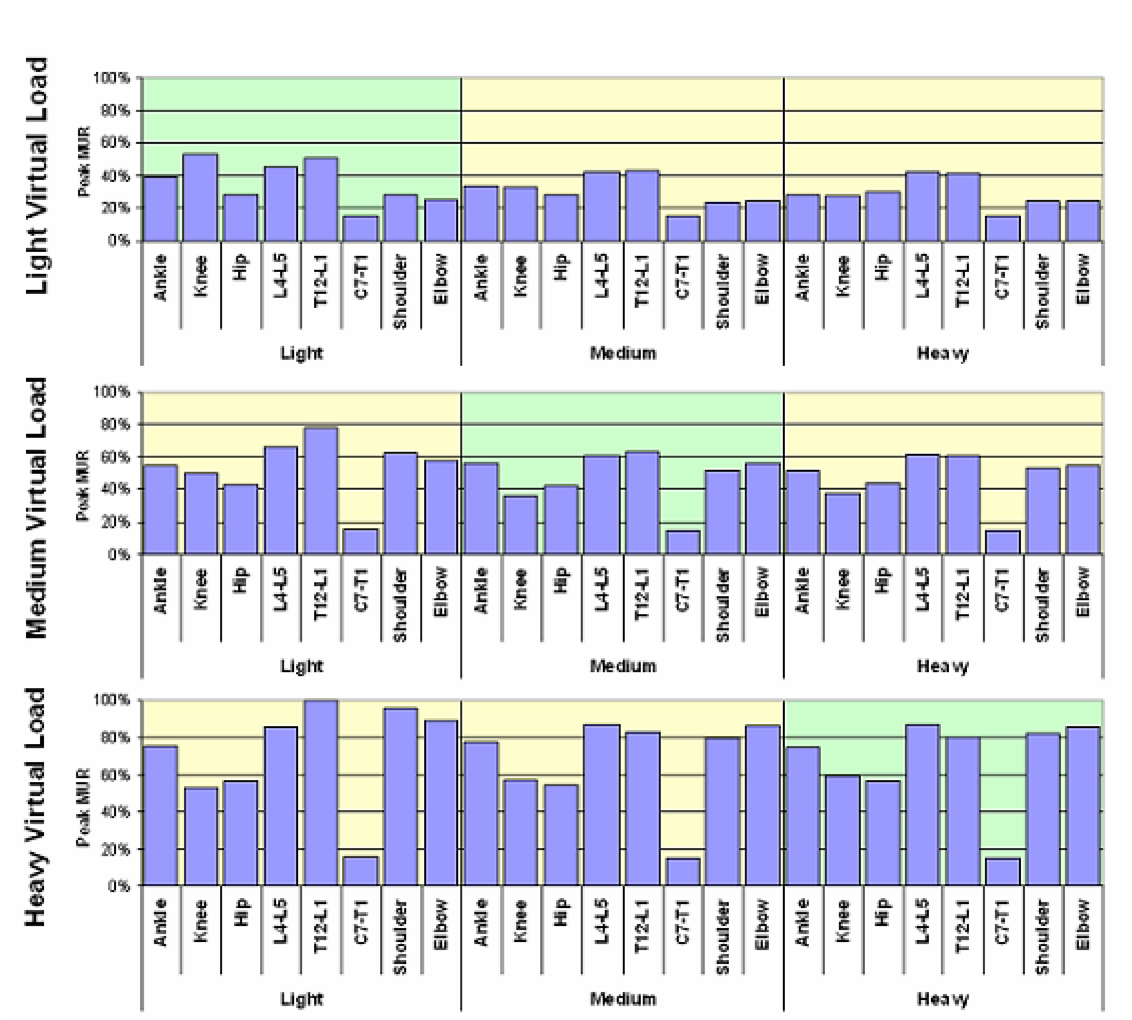


Figure 5.11. Peak Muscular Utilization Ratio (MUR) levels of each joint in all nine matrix scenarios. Peak MUR level expresses the maximum joint torque applied during the simulation as a percentage of joint strength. Higher MUR levels are observed as the model lifts heavier loads. Generally, joint peak MUR levels did not vary greatly as the model tried to follow different experimental kinematics. However, the T12-L1 joint in the vHaL scenario was the only joint out of all the scenarios to undergo maximum exertion.

and heavy experimental kinematics yielded nearly identical peak MUR levels. However, when the model was tasked with matching light experimental kinematics the peak MUR levels were slightly higher for the L4-L5, T12-L1, shoulder, and elbow joints. In the vHaL scenario, the T12-L1 joint was the only joint out of all the scenarios to apply a peak torque at 100% of its capacity. Conversely, the peak MUR level of the vHaH scenario for the T12-L1 joint was only 80%. Overall, these results suggest that all three of the subject's lifting strategies required similar levels of strength, but if the subject had chosen to adopt his light weight lifting strategy to lift the heavy weight, he might have been more prone to injuring his back.

Energy Efficiency

The efficiency of each simulated scenario using the original matrix was examined through calculations of muscle work and muscular "effort" in Equations 4.2 and 4.4, respectively. These properties were examined to determine if people might subconsciously adopt specific lifting techniques in order to conserve energy. Results from the calculations of muscle work and muscular "effort" are provided in Figure 5.12. As expected, work and "effort" both increased significantly as the model lifted heavier virtual loads. For a given virtual load, there was a slight but consistent trend that the work was reduced as the model followed heavier experimental kinematics. Therefore, from a work standpoint, the subject adopted more efficient kinematics as he lifted heavier weights.

The efficiency results estimated from muscular "effort" were somewhat inconclusive. There was a very slight tendency for the effort to increase as the model tracked heavier experimental kinematics, but this pattern was not consistent when the model lifted the heavy virtual load. It is also interesting that the faint patterns of muscular "effort" contradict the efficiency trends of muscle work.

Based on Equation 4.4, "effort" was expected to increase as the model tracked heavier experimental kinematics because the lift duration of these kinematics was

A.

B.

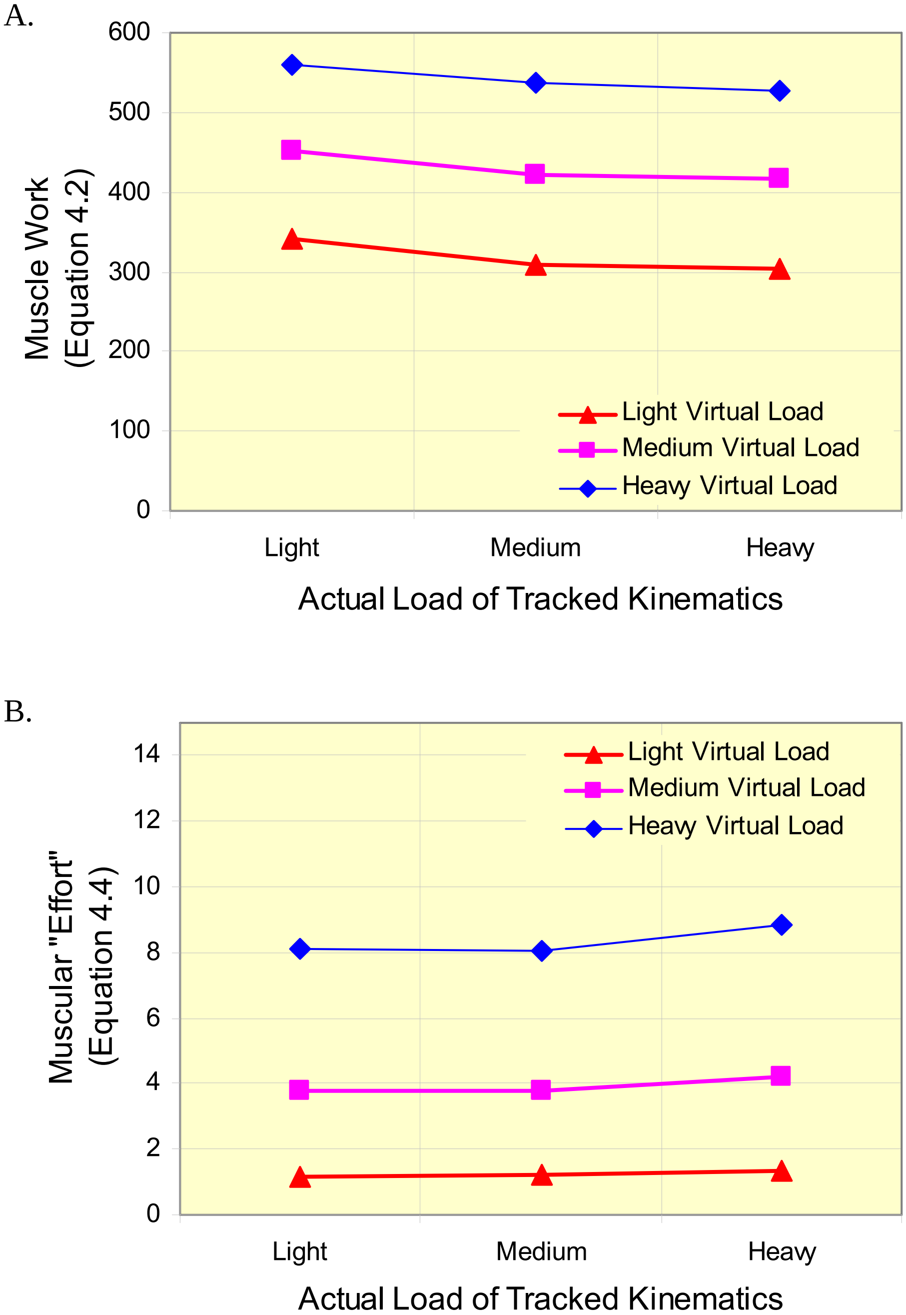


Figure 5.12. (A) Mechanical work calculated from Equation 4.2, and (B) muscular "effort" estimated from Equation 4.4. For a given virtual load, work expended decreased when tracking kinematics of the subject lifting the heavier loads. No clear pattern was observed for "effort".

longer. However, a strong relationship of this nature was not observed, which implies the heavier experimental kinematics (for a given virtual load) must have required lower joint torques on average in order to provide roughly the same muscular "effort" value.

CHAPTER SIX

Discussion

Studies have reported that lifting kinematics change predictably in response to alterations in load [12, 15-22]. To test whether these kinematics trends are intrinsic or voluntary, a computer model was developed to simulate lifting in the sagittal plane based on physical parameters and motion data from a human subject. The model was constrained to obey the laws of physics, maintain stability, and comply with the limitations of joint strength. A matrix of nine simulated scenarios was created such that for each case, the model lifted one of 3 virtual loads, while attempting to duplicate the experimental kinematics of the subject lifting his actual load. For three scenarios, the virtual and actual loads were identical. For the remaining six scenarios, the kinematics being tracked by the model came from the subject lifting an actual load that was different than the virtual load specified to the model.

Results from the simulations revealed that none of the matrix scenarios exhibited any substantial deviation from the experimental kinematics. Since the laws of physics and other constraints on the model did not cause the model to deviate from the experimental kinematics as the model lifted various virtual loads, it seems that load magnitude does not specify lifting kinematics. Further investigation into the effects of balance demonstrated that the subject lifted the heavy load in a manner that offered superior stability compared to the other experimental trials. Examination of peak joint moments revealed that generally, the peak moments did not approach strength capacity and did not vary greatly as the model tracked different experimental kinematics.

Therefore, simulated kinematics were unaffected by the limitations of joint strength. The results of joint torque analysis also indicated that by modifying the lifting movement to accommodate heavier loads the subject significantly reduced the required joint moments at the lower back. This effect may have made the subject less prone to back injury. Moreover, there was some evidence suggesting that the lifting strategies adopted by the subject to lift heavier loads were more efficient than those adopted during experimental trials with lighter loads.

From these findings, one may conclude that people are able to voluntarily manipulate lifting kinematics independent of load and do so for a variety of underlying reasons. The results could have significant implications regarding the impact of load on prescribed lifting technique. For example, it seems that lifters can perform prescribed techniques to possibly prevent LBP. Additionally, the findings may be beneficial for perception studies that seek to determine the factors people observe when viewing lifting movements.

The Debate over Impact of Load on Lifting Kinematics and Perception

Many researchers have reported that as people lift heavier loads they tend to consistently adopt specific kinematic patterns [12, 15-22]. Some debate has ensued [17, 25-29] regarding whether these kinematic tendencies are unavoidable functions of load or if people voluntarily assume the motion alterations according to some underlying motivations.

Runeson and Frykholm's KSD Principles

Runeson is one of the leading advocates supporting the hypothesis that lifting kinematics are inherently linked to the load being lifted [26-28]. Runeson and Frykholm performed experiments in which actors were instructed to lift boxes in a deceptive manner so that observers would be misled with respect to the actual weight of the box. Observers viewed point light displays of markers attached to the actors' joints so that only kinematic data would be available to the viewers. No specific instructions were given to the actors regarding which kinematic variables to manipulate in order to fake the movements. Not only were the observers able to correctly identify the intended box weights, but also they were able to distinguish the true weights of the boxes [27]. Runeson later concluded that the kinematics alone contained information that specified the difference between true and faked acts [26]. He asserted that observers inferred the box weight, which is a kinetic property, from only the actors' kinematics.

Additionally, Runeson and Frykholm proposed the KSD principles which states that kinematics directly specifies dynamics. According to these principles, observers cannot be deceived by the weight lifted because (1) changes in the kinematic pattern directly reflect the weight being lifted, and (2) it is impossible for lifters to manipulate all the kinematics to ensure that the motion required to lift another weight is matched exactly [29]. According to the KSD principles, all the dynamics of lifting are interrelated. If the load is altered, then the changes will propagate to other dependent kinematics, and the observer will be able to determine the actual weight from the shift in kinematics.

Other Perspectives on KSD Theory

Other researchers have challenged the KSD principles proposed by Runeson and Frykholm. Gilden and Proffitt contested that human perception of movement is based on heuristics or logical guidelines that observers use to make sense of the dynamics involved in a system. They suggested that in complex systems such as lifting, the kinetic effects are not intuitive, so observers cannot be expected to accurately extract dynamic properties (e.g., the load being lifted) [31]. Gilden and Proffitt identified weaknesses in the weight lifting study described by Runeson and Frykholm and attempted to discredit the KSD principles. Primarily, (1) it is unclear what information in the kinematics was used by the observers to make judgments, and (2) the constraints imposed by the experimental designs were not well defined [31]. The main complaint against experimental designs in Runeson and Frykholm's study was that the actors were not specifically instructed on how to perform a deceptive lift. Shim pointed out several factors that might have caused the actors to perform the deceptive movement in a poorly coordinated fashion and thus permitted observers to identify both the intended and actual weights being lifted:

> First, a deceptive movement would have a kinematic trajectory (including velocity characteristics) that is unnatural to the actor or the activity. Second, all action involves compensatory movements throughout the body that serve to maintain an appropriately balanced posture. Third, departure from natural movements also entails departure from their inherent economy. Fourth, separate control over the kinematics of interpolated joints (e.g., wrists and elbows) may not be upheld while one's attention is engaged in doing work with the hands. [29]

In another perception study, Shim et al. performed experiments similar to those of Runeson and Frykholm in which test subjects were asked to observe an actor lifting a weight and infer both the intended and actual weight. The actor was instructed to lift the weight more slowly than normal in order to convey that the weight was heavy and to lift at a faster pace to fake that the load was light. Contrary to the findings of Runeson and Frykholm, subjects were able to determine the intended weight correctly, but could not identify the actual weight [29]. These results indicated that weight is not directly perceived through changes in kinematic pattern and lifters can manipulate the lifting kinematics required to match those generated by a different weight.

According to Zhang et al., "one problem often encountered in understanding as well as modeling human movement is that there are an infinite number of possibilities to determine a posture due to excessive degrees of freedom (DOF[s]) possessed by the human body [39]." This dilemma is often labeled kinematic redundancy. Kinematic redundancy applies to the lifting movement because there are an infinite number of ways that a lift can be performed. Because of the excessive DOFs present in the lifting movement, it is too difficult for people to automatically conceptualize which lifting patterns, if any, might be directly related to load. However, these types of observations are much easier to identify in simple systems with only one DOF. For example, in the isolated bicep curl, it is apparent that a person with sufficient strength to lift a dumbbell would be physically able to lift a lighter dumbbell in the exact same manner. For this situation, if the upper arm is constrained and only the forearm is allowed to move, the only parameters needed to describe the movement are the angular kinematics of the elbow. Provided that the subject performing the curl has sufficient strength, the laws of physics would not prohibit kinematics observed from curling another load. It is logical that the reflections from this simple exercise might be extended to imply that load magnitude does not dictate kinematic trends in complex motions such as lifting.

Bingham recognized the difficulty in isolating kinematic variables in complex whole-body actions and sought to isolate velocity over position information in weight perception for the same bicep curl example discussed above. The study revealed that for light and medium weights, the velocity patterns did not vary significantly, and the observers had difficulty in determining the weights. However, when the heavy weight was curled, there was a large drop in angular velocity due to strength limitations, and the subjects were able to identify the heavy weight [25]. Bingham's results seem to indicate that, for the situation with only a single degree of freedom, load magnitude did not dictate specific kinematics.

Scholz argued that some dynamics involved in the lifting movement are inherently dependent on the load, while other coordination aspects may be attributed to intentional influences [17]. He believed that, on the surface, it might seem best to examine lifting in an unconstrained fashion (i.e., freestyle lift) in order to best observe the natural coordination and dynamics of lifting, but he noted several complicating factors to this approach. Scholz states:

> On one hand, lifting has received much public attention regarding the importance of lifting safely. On the other hand, individuals differ in their regard for such publicity and in their previous experiences with lifting. Therefore, it is difficult to ascertain what biases about lifting a subject brings to the experiment. As a result, it would be extremely difficult to separate the effects of intrinsic from intentional dynamics. We chose to specify the pattern to be used to minimize differences in intentional influences on the lifting dynamics. The changes in coordination found in this experiment can, therefore, more likely be attributed to the influence of intrinsic dynamics on the coordination of this task. [17]

Essentially, Scholz recognized that if lifting technique is not specified, different subjects will naturally adopt their own, unpredictable lifting strategies. Since Scholz's investigation sought to isolate the generalized dynamics and coordination of lifting via purely experimental procedures, he was required to impose a specific lifting technique on his subjects. By prescribing the lifting pattern for the subjects to use, Scholz was restricted from examining the manner by which people naturally perform manual lifts. Additionally, Scholz admitted that specifying the lifting technique could not fully isolate the intrinsic (i.e., independent of voluntary influences) dynamics on the coordination of the lift but will hopefully reduce the intentional influences.

Scholz's dilemma illustrates that it is unrealistic to expect all subjects to perform lifts in exactly the same manner each time. Even when the lifting technique is prescribed, subjects can be expected to exhibit some deviation from the desired motion because the subjects' brains do not provide precise quantitative control over coordination. These inherent shortcomings in human experimental lifting trials highlight the advantages of the use of computer-simulated biomechanical models. Unlike human experimental data collection methods, computer simulations are extremely consistent, generating the same simulated results repeatedly.

Findings from the Current Study

In the current study, the advantages of computer models were used to attempt to isolate potential intrinsic dynamics that might be directly specified by the load. To accomplish this goal, the model was required to lift a particular virtual load while attempting to match the experimental kinematics of the subject as he lifted an actual load. The reasoning for this method was: If the model was found to deviate from the experimental kinematics, the results would suggest that the load inherently specifies the kinematics. Alternatively, if the model was able to match the desired motion regardless of the magnitude of the load, these findings would imply that people are physically capable of manipulating kinematics despite the inertial effects of the load and for underlying motivations, elect to perform the lift with the commonly recognized changes in technique.

Results indicated that for every scenario in the original matrix (see Figure 4.1), inertial effects of the load did *not* prevent the model from simulating essentially identical kinematics as those exhibited by the subject as he lifted alternate loads than were specified to the model. These findings indicate that movement is not dictated by the load.

To further substantiate the model's findings, tighter constraints were imposed on the model's requirements to maintain stability, and a new set of simulated-scenarios called the balance matrix was created. The new stability criteria were significantly more prohibitive than traditional studies that have examined the effect of balance on lifting [33, 36, 37]. The results of the balance matrix again revealed that load did *not* necessitate altered kinematics unless the model was placed in a position of extreme vulnerability to losing balance. These cases only arose when the model was given the heavy load and was expected to follow the medium and light kinematics. For these two balance matrix scenarios, the model was still able to generate the desired experimental kinematics until the load was lifted far away from the body to be placed on the weight rack. Only at this point did the model have to adjust slightly with a maximum joint deviation of 3.6 cm by shifting the COM heel-ward in order to maintain appropriate stability. Therefore, the evidence provided by the model strongly suggests that load does not directly designate kinematics or vice versa.

The findings of the current study contradict the KSD principles proposed by Runeson and Frykholm [26-28] that "observers can directly perceive the kinetic property (ie mass or weight) from the kinematics (eg displacement, velocity, and acceleration of the joints) [30]." According to the biomechanical model, neither physical laws, nor strength, nor stability limitations dictate changes in kinematics in response to load magnitude. Analysis of these findings only suggests changes in kinematics due to load are not mandatory, but lifters may voluntarily make adjustments in technique without consciously realizing it or without knowing how to avoid the changes. Therefore, the findings of the current study did not dispute theories related to human perception, but it seems that for the case of lifting, kinematics do not fully specify the dynamics as the KSD theory proposes.

A final point to note is that it is common for lifters to demonstrate atypical kinematic patterns as load is altered. Clearly, many sources have published results expressing similar patterns that subjects often adopt in response to load [12, 15-22], but an equally important, yet often overlooked, observation is that lifters are able to willingly defy these patterns. Examples of published studies reporting these types of results include Lindbeck and Kjellberg and Burgess-Limerick et al. who identified some subjects who did not demonstrate an increasing lag between the knee and hip as load was increased [9, 12]. In a study on lifting speed, researchers reported that the weight of the load did not affect lift velocity or duration [56]. Similarly, the current study's subject also demonstrated atypical trends for some of the experimental trials. In order to study lifting movements thought to be representative of the general lifting population, each of the trials was scrutinized to select representative trials for each load category (i.e., light, medium, and heavy) that did follow the trends reported in the literature. This observation that atypical kinematic patterns are common is important because it illustrates that, as long as the lifter is sufficiently strong and in a stable position, the lifting kinematics are largely independent of the load. According to this notion, physical laws do not fully dictate kinematic trends such that the trends will always be present and consistent.

Motivation for Voluntarily Movement Alterations in Response to Load

Once the results of the model revealed that the tendency of lifters to assume specific kinematic patterns in response to load magnitude was nonmandatory, steps were taken to examine voluntary factors that might cause people to lift according to the documented trends.

Balance

For the original matrix, there were no difficulties among any of the scenarios for the model to adhere to the commonly reported conditions required to maintain appropriate stability [5, 33, 36, 37]. The implication of this finding is that balance requirements do not dictate changes in lifting technique in response to altered load magnitude.

To further examine this assessment, the balance matrix simulations were performed to observe the impact of restricting stability according to extremely strict specifications. Typical studies that investigate the effect of balance on lifting only require that the COP, which is usually defined simply as the total COM, remains within the BOS, which extends from the end of the toe to the heel [33, 36, 37]. The balance matrix, however, constrained the simulations to maintain COP within a reduced BOS that was defined according to the COP limits that the subject adhered to for the heavy load.

While the balance matrix results were not intended to examine the assumptions of how BOS is commonly defined, they did suggest an interesting finding. Both force platform data and the model agreed that, for all three of the actual experimental trials, the subject maintained stability by keeping his COP between the heel and within 80% of the distance from the ankle to the toe. This distance reflects the approximate length from the ankle to the balls of the feet. These findings may imply that, in order for adequate stability to be maintained, the COP must remain between the heel and a point closer to the ankle than the toe (i.e., perhaps the balls of the feet).

Interestingly, the balance matrix simulations revealed that except for two out of the nine simulations, the more rigid stability conditions still did not prohibit the model from matching the target experimental kinematics. Only two scenarios violated the reduced BOS, vHaL and vHaM, and for these the experimental kinematics were still tracked perfectly by the model until the point at which the heavy weight was to be placed on the rack. Only at this extreme moment of vulnerability for the model to lose balance were the desired kinematics compromised slightly (i.e., maximum joint deviation of a few centimeters) in order to maintain appropriate stability. Therefore, the kinematics that the subject chose to lift the heavy weight offered better stability at the point during the lift in which the subject was most vulnerable to losing balance. These findings suggest that although the subject was not required to alter his kinematics for heavier loads, he may have voluntarily chosen to use more stable kinematics according to the load.

Mechanical and Muscle Work

Analysis of muscle work, calculated according to Equation 4.2, suggested that as the subject lifted heavier loads the style he used to lift the weight required less work and was, therefore, more efficient. This trend was revealed without exception among each matrix scenario, which emphasizes the possibility that people may voluntarily alter their self-chosen lift techniques for heavier loads to reduce work and, thus, energy expended as a result of the movement.

It is of interest to examine how the subject's altered kinematics reduced work. As explained in Chapter 4, the muscle work required for a task is equivalent to the mechanical work performed on the total COM (i.e., combined COM of the body segments and load). Since mechanical work is also equivalent to the gravitational force acting on the total COM multiplied by the displacement of the COM, the only way for work to be reduced for a given virtual load was to reduce the displacement of the COM. Furthermore, since the weight rack required the initial and final positions of the load to be the same for each trial, work was reduced for heavier experimental kinematics by moving the body segments differently in such a way that the vertical COM position was initially higher, and thus moved a shorter distance.

Other studies have pointed out that this pattern of reducing work by elevating initial COM position is typical of the transition from a squat lift to a stoop technique [8, 11]. For the traditional squat method, initially the knees are bent so that the legs will provide the majority of the work rather than the back. The back is relatively straight but is inclined forward to grasp the weight. This position automatically lowers the total COM relative to the stoop technique, which elevates the body by maintaining straighter knees (see Figure 6.1). Researchers have also observed that as the load is increased, lifters tend to employ more of a stoop technique than for lighter loads [11, 15, 19]. In the current study, the subject also adopted a lifting technique that approached more of a stoop lift as he lifted heavier loads. This observation is apparent from both the kinematics and mechanical work calculations. Therefore, analysis of work seems to further reinforce that the selected experimental trials performed by the subject were representative of the general lifting population.

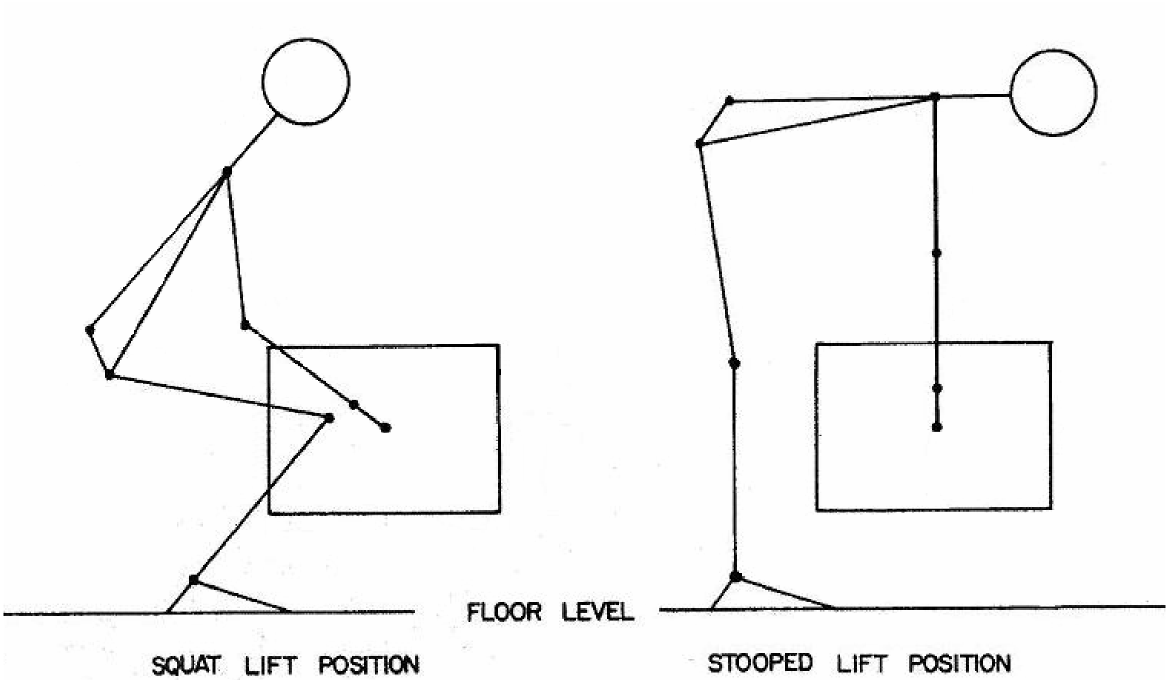


Figure 6.1. Illustration depicting two common lift techniques: stoop and squat adapted from Garg and Herrin [8]. In the squat technique, the legs perform most of the work, whereas the stoop lifts primarily with the back. The squat approach is more commonly prescribed as the safest technique, but increasing research supports advantages of the stoop lift.

The stoop lift has been suggested to be superior to the squat in terms of balance, knee clearance, the level of fatigue experienced, and metabolic energy expended [4, 8, 10, 11]. Garg and Herrin report, “The metabolic costs favor the stooped posture in terms of lower heart rates and metabolic energy expenditure rates for lifting a given load. The squat posture contributes more to physical fatigue than the stooped posture. For the same level of physical fatigue, a greater amount of mechanical work can be accomplished if the stooped posture is employed [8].” These advantages of the stoop lift may encourage people to adjust lifting technique to more closely resemble the stoop as load is increased.

Muscular “Effort”

Calculations of muscular “effort” according to Equation 4.4 did not demonstrate obvious improved efficiency as the simulations tracked the subject lifting heavier loads. The “effort” values did not vary greatly according to the different kinematics adopted by the subject. Contrary to work efficiency, there was a subtle tendency for the efficiency estimated by muscular “effort” to worsen as the subject lifted heavier loads. In terms of efficiency, Gagnon and Smyth note that the calculation of muscular “effort” according to the ratio of joint moment to joint strength is a procedure that should only be applied to supplement the information on work energy variables [55]. Therefore, less credit should be attached to muscular “effort” efficiency results than to direct work-related measures. However, by considering the way muscular “effort” was derived, insight was gained that may potentially explain why subjects tend to predictably adjust their lifting strategies to accommodate heavier loads.

Since muscular “effort” was calculated by integrating moment ratios over the time duration of the lift, longer trials were expected to provide larger “effort” values. A study on lifting speed by Lin et al. supported this expectation by showing that muscular “effort”, as calculated by Equation 4.4, was reduced for faster lifts [56]. For the current study, the heavier experimental trials had significantly longer durations (see Figure 5.2), yet the muscular “effort” of simulations tracking these kinematics did not yield larger values as anticipated. Since a strong relationship of this nature was not observed, it appears that, in a metabolic sense, the subject adopted lifting strategies for heavier loads that compensated for longer lift duration by reducing the required moments about the joints. This supposition is to some extent supported by the peak MUR levels given by

Figure 5.11 because the light experimental kinematics required slightly higher peak torques than the medium and heavy kinematics. However, analysis of Figure 5.11 does not fully interpret the results of muscular “effort” because MUR levels were based on peak torques, whereas muscular “effort” was calculated from the integration of instantaneous joint torques. Therefore, it is possible that muscular “effort” may have yielded a lower value for the heavier experimental kinematics than was anticipated (from the longer durations) due to slight reductions in many required torques rather than a large reduction in the peak torque of a given joint. This effect could be present, yet indistinguishable from the peak MUR levels of Figure 5.11.

The above argument brings up several interesting points. While straightforward analysis of the muscular “effort” results did not immediately reveal trends that efficiency was improved as the model adopted experimental kinematics from the subject lifting heavier wieghts, reflection over the “effort” results identifies that the heavier experimental kinematics required the model to generate lower joint moments on average. These results might have otherwise been overlooked because they were not obvious from analysis of peak MUR levels, which only investigated peak joint torques. The significance is that lifters may choose the common technique transitions for heavier loads because it is less metabolically taxing to take longer to perform the lift in order to reduce joint torque on average. For light loads, average torques may be small enough that lifters are willing to increase the torques so that they can finish the lift sooner. This conclusion is also supported by Buseck who found that the joint moment/load relationship is significantly influenced by lifting posture and speed [57].

Metabolic Energy Expenditure

While mechanical work and muscular "effort" calculations may imply properties of efficiency, ultimately they fall far short of accurately describing the true metabolic efficiency of a movement. As Burdett et al. pointed out, "Unfortunately, neither mechanical work nor joint moments are exact measures of energy consumption. Factors such as different muscle fiber types, co-contractions of antagonistic muscles, elastic storage of energy in muscles and ligaments, and isometric contractions of muscles are not reflected in these measurements [58]." Theoretically, the ideal measure of efficiency is metabolic energy expenditure, which takes into account the above factors as well as muscle fatigue and exertion during static postures. However, it is extremely difficult to accurately quantify metabolic energy expenditure, especially for short movement durations. Energy expenditure measurements also depend on parameters such as oxygen consumption and specific muscle properties. Therefore, like other lifting simulation studies, energy consumption could not be calculated by the model used in this investigation. However, it is likely that if metabolic energy expenditure could be evaluated, accurate energy efficiency quantifications of the different lifting strategies would be very insightful.

Injury

It may be possible that people unconsciously adjust their technique as load increases in order to avoid straining more vulnerable parts of the body (e.g., the back). Many have suggested that the technique spontaneously chosen by the subject may be the least likely to lead to injury [8, 12]. Anderson believes "it is safer to allow workers to use their own common sense and muscle sense than to teach them new drills in performing certain jobs in which a series of predetermined positions must be consciously assumed [59].” This statement is supported by observations of manual materials handling workers that the squat technique is rarely used to lift heavy loads [8].

The current study also illustrated that, for the heavy virtual load, the subject’s self-selected lifting strategy required substantially lower peak torques than were necessary for the model to match the light experimental kinematics. In fact, the light experimental kinematics required the model to exert the T12-L1 joint (i.e., upper lumbar joint in lower back) at 100% of its capacity compared to the heavy experimental kinematics which required a peak moment at only 80% of the potential joint strength for the same joint. According to the model, the subject could have lifted the heavy load according to this alternate set of kinematics, but if he had done so, he would have had to use maximal lower back torque, putting him at greater risk of injury. Therefore, the documented kinematic patterns associated with changes in load might be chosen by lifters in order to reduce the likelihood of pain or damage.

Lifting Technique

As noted above, lifting technique, which includes both posture and speed, plays a key role in determining the joint loads and moments experienced during a lift. This, in turn, affects the lifter’s likelihood of injury. Many have acknowledged that, by prescribing lifting technique injuries might be prevented [4, 8, 9, 10, 11, 12, 15]. However, there is much debate over which lifting technique, if any, should be prescribed to lifters. The most commonly described lifting strategies are the squat, stoop, and semisquat (see Figure 6.1). The semisquat refers to an intermediate posture between the squat and the stoop. According to Heiss et al., “Persons who are allowed to use a freestyle technique (ie, the technique that is most comfortable for them) adopt the semisquat technique [11].” This general comment also held true for the freestyle lift chosen by the subject in the current study.

Shipplein et al. stated, “A prescribed lifting technique might not be possible due to the amount of weight lifted and/or the ability of the subject [15].” The results of the current study demonstrated that the model was able to generate different sets of experimental kinematics despite the effects of the load. Additionally, analysis of each joint’s peak moments revealed similar peak joint torque levels for each set of experimental kinematics that the simulations tried to match. The implication of this finding was that strength was not a limiting factor that necessitates specific kinematics. Therefore, simulation results suggest that the subject would not have been limited by strength or any other factor to prevent him from adopting prescribed movements.

However, the experimental kinematics examined in the current study did not represent substantially different movements. The distinction is that the subject chose to use fairly similar semisquat techniques to lift the light, medium, and heavy loads. It is not clear what results the simulations would have provided if the model was tasked with matching kinematics of a pure squat lift or pure stoop lift. Thus, it might be insightful for a subsequent study to simulate the impact of load on widely different lifting kinematics. The new study might better elucidate details and factors that could be important for defining the optimal lifting technique for the majority of the population.

Justification of Methods

Modeling

The combined elements of biomechanical modeling and computer simulation offer a widely accepted approach to learning about the behavior of human movement. Computer modeling tends to be extremely robust in its ability to describe interactions between bodies according to the laws of physics. Additionally, modeling facilitates calculation of variables that cannot be measured or assessed directly. In the current study, manual lifting was analyzed by implementing modeling and simulation techniques that were demonstrated successfully in similar investigations. Considerable complexity was avoided by modeling lifting in only two dimensions as other lifting studies have done. Two-dimensional lifting models are the norm because lifting naturally occurs almost exclusively in the sagittal plane. Additionally, the experimental lifting protocol restricted the subject, on whom the model was based, from using non-sagittal movements. To seek to model the lifting motion as accurately as possible, the model developed for the current study added several significant features beyond those used in previous studies.

One important feature of the model was the addition of the pelvis, abdomen, thorax, and head segments, which are often neglected in typical lifting studies [5, 13, 32, 34-37]. The partitioning of the trunk into three segments was essential to representing the lower back, which plays a key role in elevating the upper body and load. The primary advantages of representing the trunk as three separate segments were: (1) critically important coordination movements could be observed between back segments, (2) moments could be calculated to indicate low-back exertion, (3) better inertial properties could be defined to provide more realistic dynamics of the model, and (4) inverse dynamics analysis errors could be reduced [46]. Similarly, the head was needed to satisfy stringent requirements for alterations in momentum and COM that may have been produced as a result of head movement.

There were also features of the current study that might have been improved. As noted in Chapter 3, average male position-dependent strength data was obtained for each joint from various documented sources in the literature [48-54]. It has been widely reported that joint strength capability changes not only as a function of position but also in response to joint angular velocity. Like other lifting model investigations, in the current study effects of velocity on joint strength were neglected. In actuality, joint strengths are reduced slightly under concentric contractions as in lifting, but these effects were considered to be insignificant because of the low velocities experienced during lifting (e.g., compared to ballistic movements such as jumping). Additionally, position-dependent isokinetic strength documentation was not available for each joint of interest. Further efforts to obtain isokinetic strength profiles would be beneficial for studies seeking to model joint strength with a high degree of certainty.

Modeling of strength might also have been improved by scaling joint torque capacities specifically for the subject. In the current study, the strength sources were assumed to be representative of the current subject. This assumption may not have been completely grounded since alternative studies, that also reported “average” male joint strengths, provided substantially different results. The joint strength data of other investigations may have varied due to differences in subjects and/or experimental measurement methods. However, in the current study, inverse dynamics analysis of the heavy experimental lift performed by the subject seemed to indicate that the joint strengths provided to the model were representative of the subject. Figure 5.11 reveals that the highest peak MUR levels in the vHaH scenario were just over 80% of the subject's capacity. Intuitively, this validates the strength values because for that simulation, the subject was lifting approximately 80% of his MAWL.

Furthermore, the required accuracy of joint strength was found to be less important than might have been expected. Previously, strength was thought to be a potential limiting factor that might cause inherent kinematic differences in response to altering the load. Therefore, it was strongly desired to have joint strengths that perfectly matched the corresponding strengths of the subject. Simulated results, however, reported that motion is not dependent on the load but is instead chosen voluntarily. Additionally, analysis of peak joint MUR levels revealed that the model was not influenced by joint strength in regards to its ability to match the target kinematics of each matrix scenario.

Consistency Checks

Throughout the development of the model, many checks were performed to verify that the methods and results were reasonable and consistent with other data. These checks prevented anomalies from being left unresolved. Several examples of the checks follow. As explained in Chapter 3, every time inverse dynamics analysis was performed, resulting torques were tested via forward dynamics to ensure that the resulting accelerations were the same as those used to drive the inverse dynamics. Checks were also implemented to verify that the torques generated by inverse dynamics were reasonable given the published joint strength data. Computer-simulated calculations of COM and COP were validated by force platform measurements. Similarly model computations of muscle work were compared to manual calculations of mechanical work.

Furthermore, every function and routine in the custom C++ software was extensively and individually tested to ensure proper logic and implementation.

A Single Subject

Another important point to note was the use of only one subject, on whom the model was based. Aside from the logistics, the primary rationale for using a single subject related to the purpose of the investigation, which was to determine if inherent properties of the load dictate lifting kinematics. While this topic of debate has not yet been fully resolved, it clearly has a “yes” or “no” answer. Namely, “Yes”, the magnitude of the load will, by the laws of physics and other factors, influence the lifting movement despite the intentional efforts of the lifter, or “No”, kinematics are fully manipulable by the lifter and are not intrinsically dependent on the load. Therefore, it was expected that if the model was implemented properly, the correct answer would arise regardless of the number of subjects.

Additionally, it was desired to examine the commonly reported lifting trends according to kinematics demonstrated by a real subject. It was thought that an “average” lift obtained from many subjects and/or trials would not accurately represent an actual lift and would thus be less relevant for the study. It made little sense to try to isolate kinematic dependencies to load for an “average” lift. However, if the methods of the current study were repeated for more subjects on an individual basis, the expected result that kinematics are not intrinsically related to load for any subject might be confirmed, thus adding credibility to this study’s findings.

CHAPTER SEVEN

Conclusions

The model was designed to represent the anthropometry, inertia, and dynamics of the human subject. It seems that the light, medium, and heavy kinematics selected from the subject's experimental trials were representative of the commonly documented kinematic patterns that are associated with changes in load. Furthermore, an appropriate balance between simplicity and reality seems to have been conserved so that the model was properly suited to isolate potential kinematic dependencies related to load. The techniques implemented followed from the successful results of previous studies, and multiple checks were employed to ensure rational and consistent results.

For every scenario, the model was successful in its task to generate essentially identical kinematics as those exhibited by the subject's experimental trials. Even when balance constraints were specified to be far more restricting than is typically implemented in lifting simulation studies, the model was able to duplicate the desired experimental kinematics for seven of the nine scenarios. For the balance matrix, only the vHaL and vHaM scenarios required the model to compromise positions slightly in order to maintain stability at the very end of the movement when the subject was in a position of extreme vulnerability to losing balance. Thus, the model revealed that the kinematics adopted by the subject to lift the light and medium loads did not provide as much stability as the lifting technique chosen by the subject to lift the heavy load. These findings suggest that intrinsic limitations (i.e., laws of physics, strength limits, and balance) do not overly constrain the choice of lifting coordination, except possibly in cases of extreme vulnerability to balance loss.

Additional assessments of the simulations implied several motives that might influence lifters to voluntarily adopt commonly observed kinematic patterns in response to load. Calculations of work revealed that the subject used more energy-efficient lifting techniques as he picked up heavier loads. According to analysis of muscular “effort,” it seems that while heavier kinematics took longer, they required lower joint moments, on average, which might have made the heavier experimental kinematics less metabolically taxing. Conversely, the subject may have purposely lifted lighter loads faster at the cost of increasing average joint moments because the joint moments were still low enough not to be a concern. Examination of peak joint torques revealed that, by choosing to lift the heavy weight according to the heavy experimental kinematics rather than the light experimental motion, the subject reduced the peak moment of a joint in the lower back from 100% exertion to only 80%. Therefore the subject chose a lifting strategy that seems to have significantly reduced his chances of injury.

The results of the study were quite enlightening and offer insight for investigations dealing with human perception of lifting as well as low-back pain and lifting technique. A related project for future study might include a computer-simulated biomechanical model similar to this one that seeks to determine the effects of load on more widely distributed kinematics for multiple subjects. For the study, researchers might employ a greater number of subjects to be examined on an individual basis and observe effects of prescribing lift techniques such as the stoop and squat. In addition, more complex modeling methods such as muscle forces might be implemented rather than joint torques so that metabolic energy expenditure might be better analyzed. Such a study might elucidate the intrinsic factors that potentially come into play when highly different lifting techniques are prescribed, and thus, aid in the understanding of prescribed lifting techniques.

APPENDICES

APPENDIX A

Lift Speed and Duration for each Experimental Trial

Valid subject data was collected for eight experimental lifts: 3 light, 2 medium, and 3 heavy. Among these experimental trials, three were chosen, one for each load category, to be representative of the general lifting population. The representative trials were selected from the eight experimental lifts to follow documented lift velocity and duration trends. The second trial of each load was selected.

The literature commonly reports that as heavier weights are lifted, lifting speed is reduced and lift durations increase [12, 17, 18, 29]. Peak vertical wrist velocity of each experimental trial is provided in Figure A.1. Consistent with reported trends, lifting speeds of the selected trials decrease as the subject lifts heavier loads. Similarly, Figure A.2. exhibits the lift duration of each experimental trial. Lift duration increases as heavier loads are lifted for the selected representative trials.

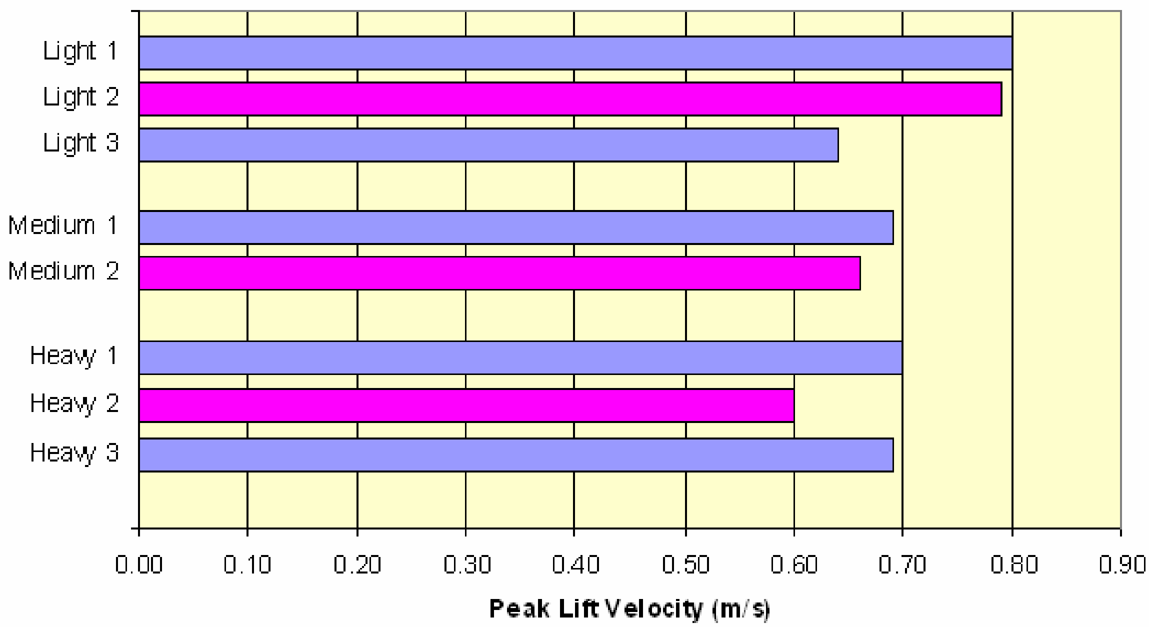


Figure A.1. Lift velocity for each experimental trial.

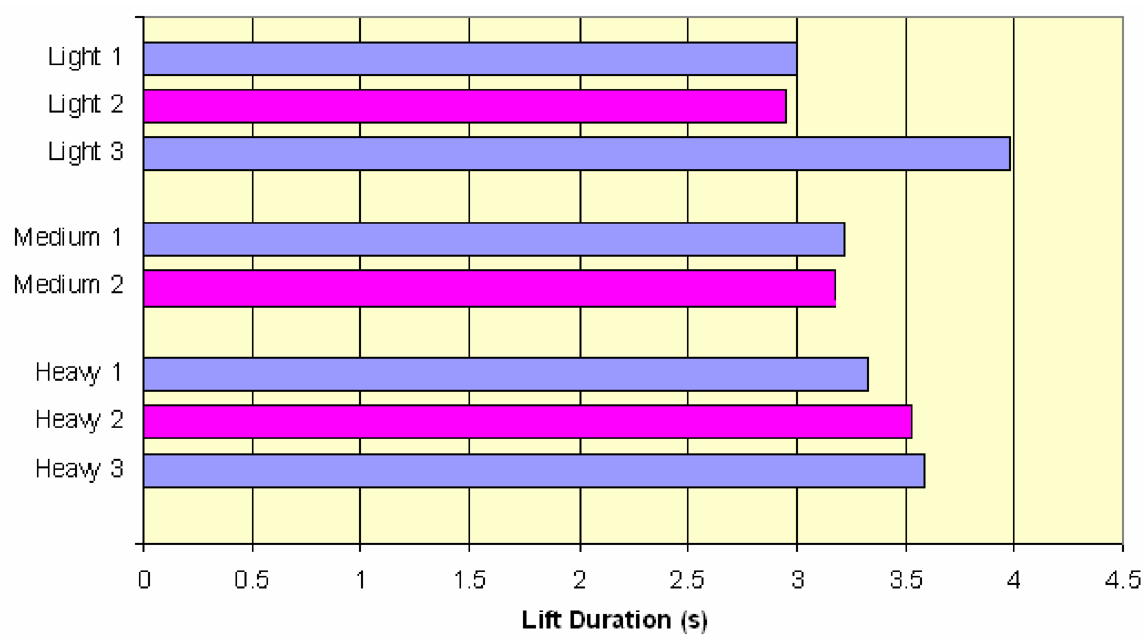


Figure A.2. Lift duration for each experimental trial.

APPENDIX B

Interjoint Coordination for each Experimental Trial

Valid subject data was collected for eight experimental lifts: 3 light, 2 medium, and 3 heavy. Among these experimental trials, three were chosen, one for each load category, to be representative of the general lifting population. The representative trials were selected from the eight experimental lifts to follow documented interjoint coordination trends. Researchers commonly reports that as heavier weights are lifted, lumbar extension occurs increasingly later than knee extension [12, 15-22].

Figures B.1-B.8 provide normalized angular position time series overplots of the knee and lumbar spine for each experimental trial. The lumbar spine angle was represented by the sum of the hip and L4-L5 joints. Both joint angles were normalized from 0 to 1, such that 1 represents greater extension. Lag between the knee and back can be observed by the joint extension difference between the joints for a given point in time. The selected representative trials were: light 2, medium 2, and heavy 2, and the desired pattern is demonstrated for these trials.

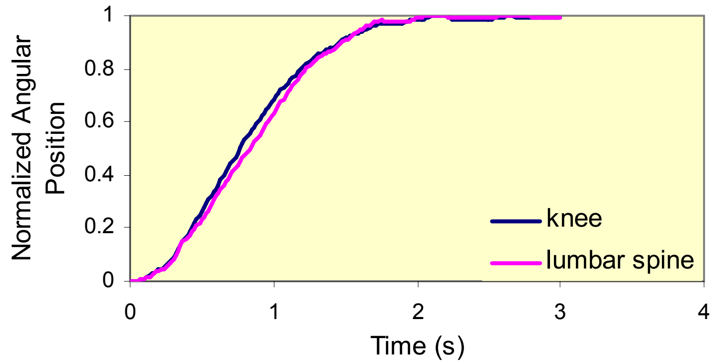


Figure B.1. Light 1 experimental trial interjoint coordination.

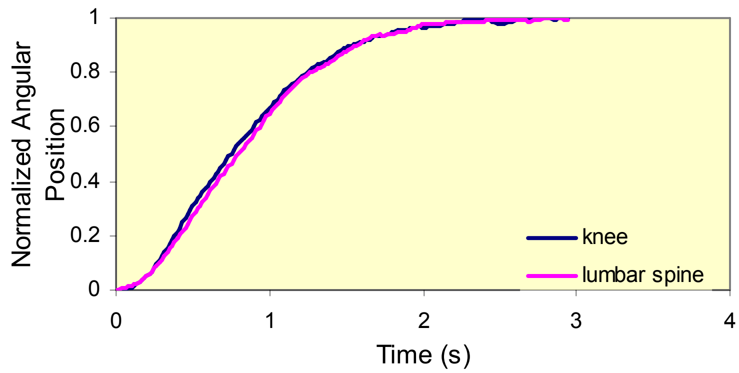


Figure B.2. Light 2 experimental trial interjoint coordination.

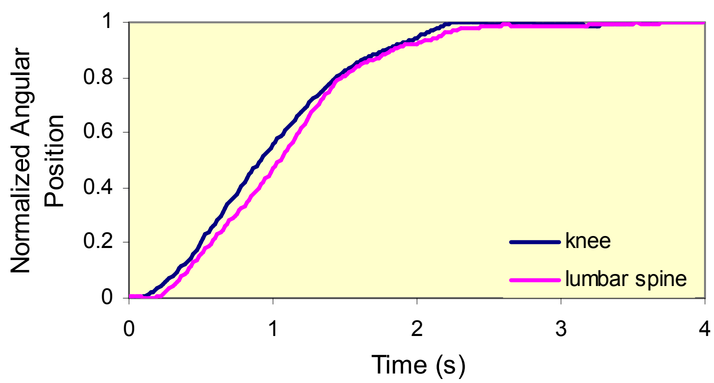


Figure B.3. Light 3 experimental trial interjoint coordination.

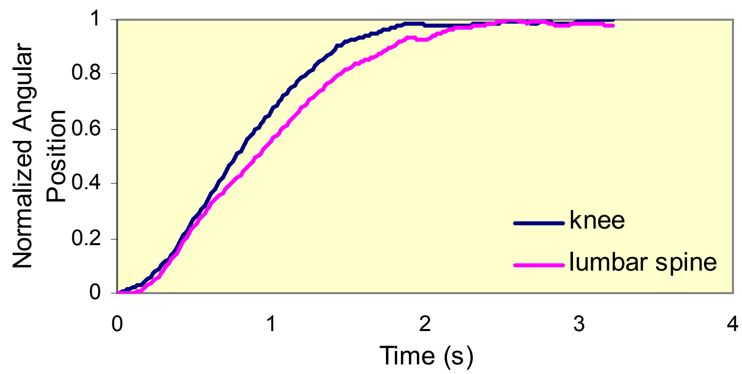


Figure B.4. Medium 1 experimental trial interjoint coordination.

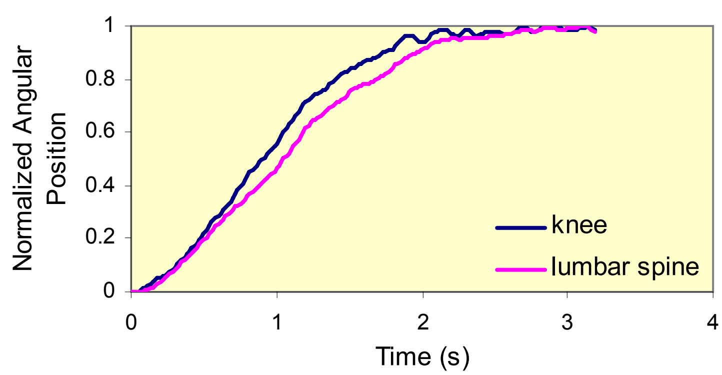


Figure B.5. Medium 2 experimental trial interjoint coordination.

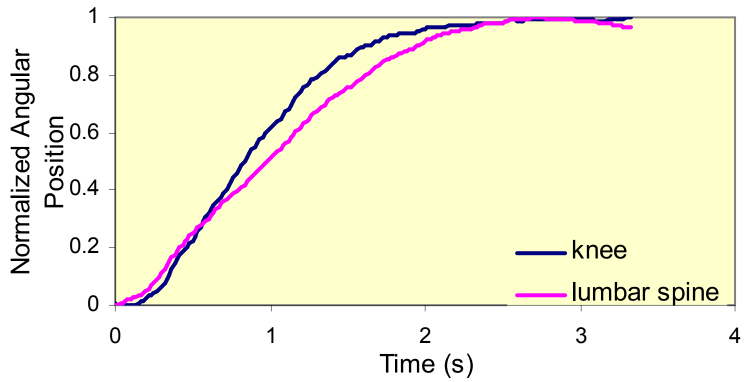


Figure B.6. Heavy 1 experimental trial interjoint coordination.

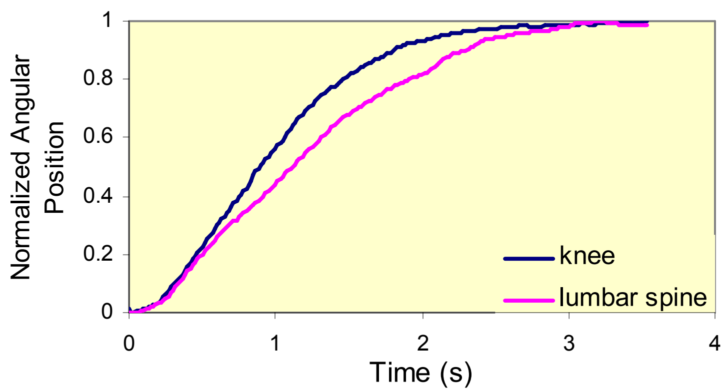


Figure B.7. Heavy 2 experimental trial interjoint coordination.

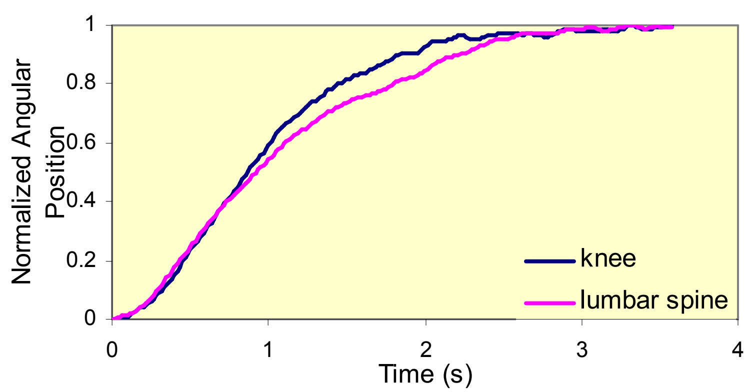


Figure B.8. Heavy 3 experimental trial interjoint coordination.

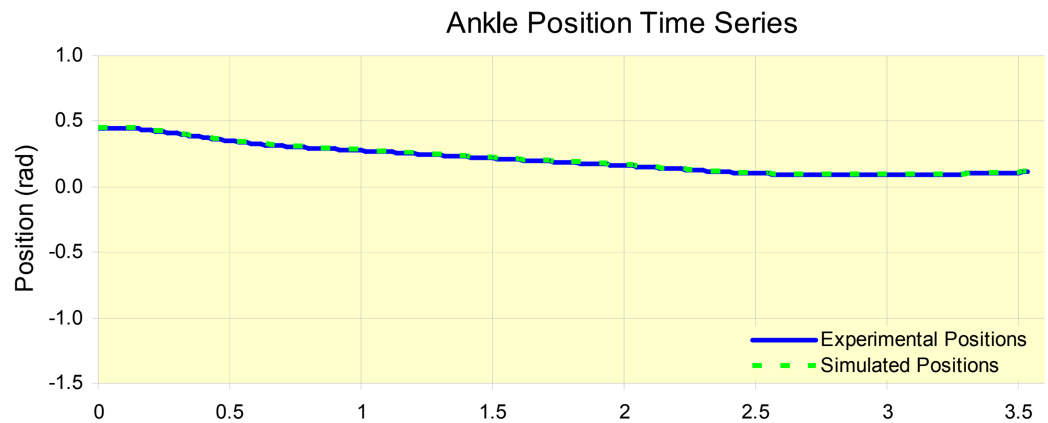
APPENDIX C

Kinematic and Kinetic Time Series for each Scenario

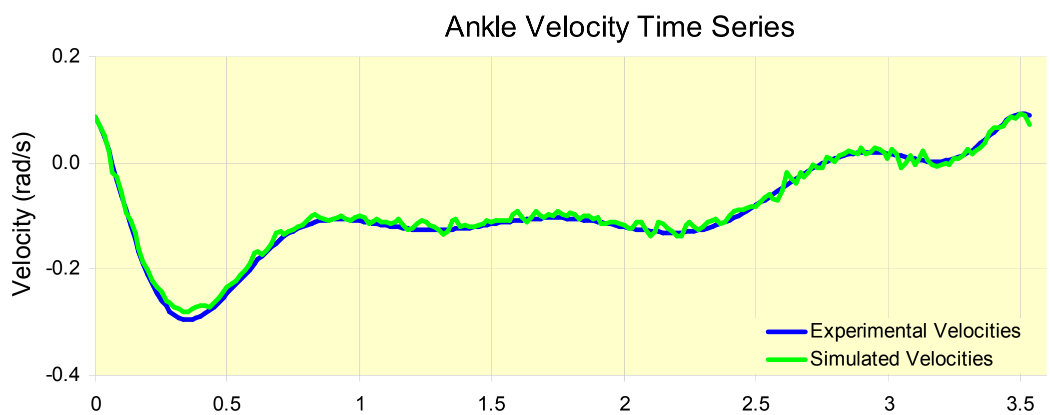
Nine scenarios were simulated to determine if physical laws would necessitate that the model deviate from the subject's experimental kinematics in order to lift a different load than the subject lifted. All nine scenarios exhibited essentially zero deviation from the desired experimental kinematics.

Plots of both simulated and experimental kinematic and kinetic data for each joint of one simulation (vMaH) are presented in Figures C.1-C.8. Others simulated scenarios were similar in their abilities to track the experimental kinematics. Figures C.1-C.8 represent the ankle, knee, hip, L4-L5, T12-L1, C7-T1, shoulder, and elbow, respectively. Each figure contains three components: (A), (B), and (C). Angular position time series plots are provided in (A), angular velocity time series plots in (B), and joint torque time series plots in (C). (C) also contains maximum strength capability under conditions of flexion and extension which vary according to joint angular position.

A.



B.



C.

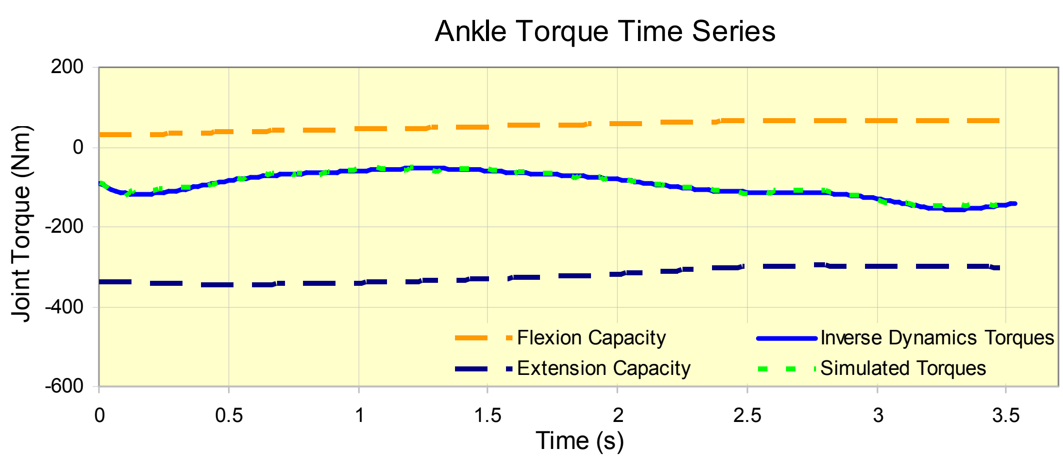


Figure C.1. Ankle joint time series.

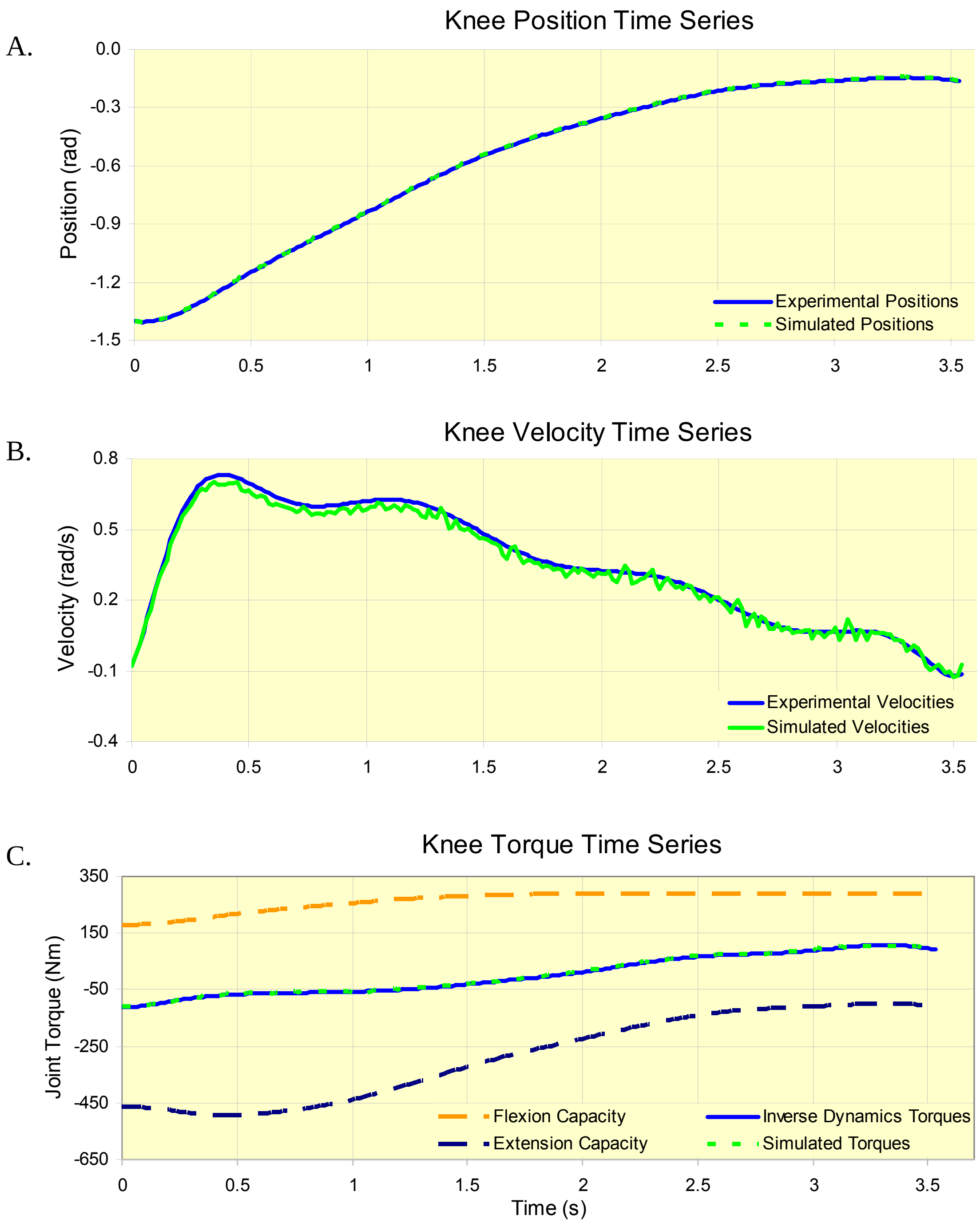


Figure C.2. Knee joint time series.

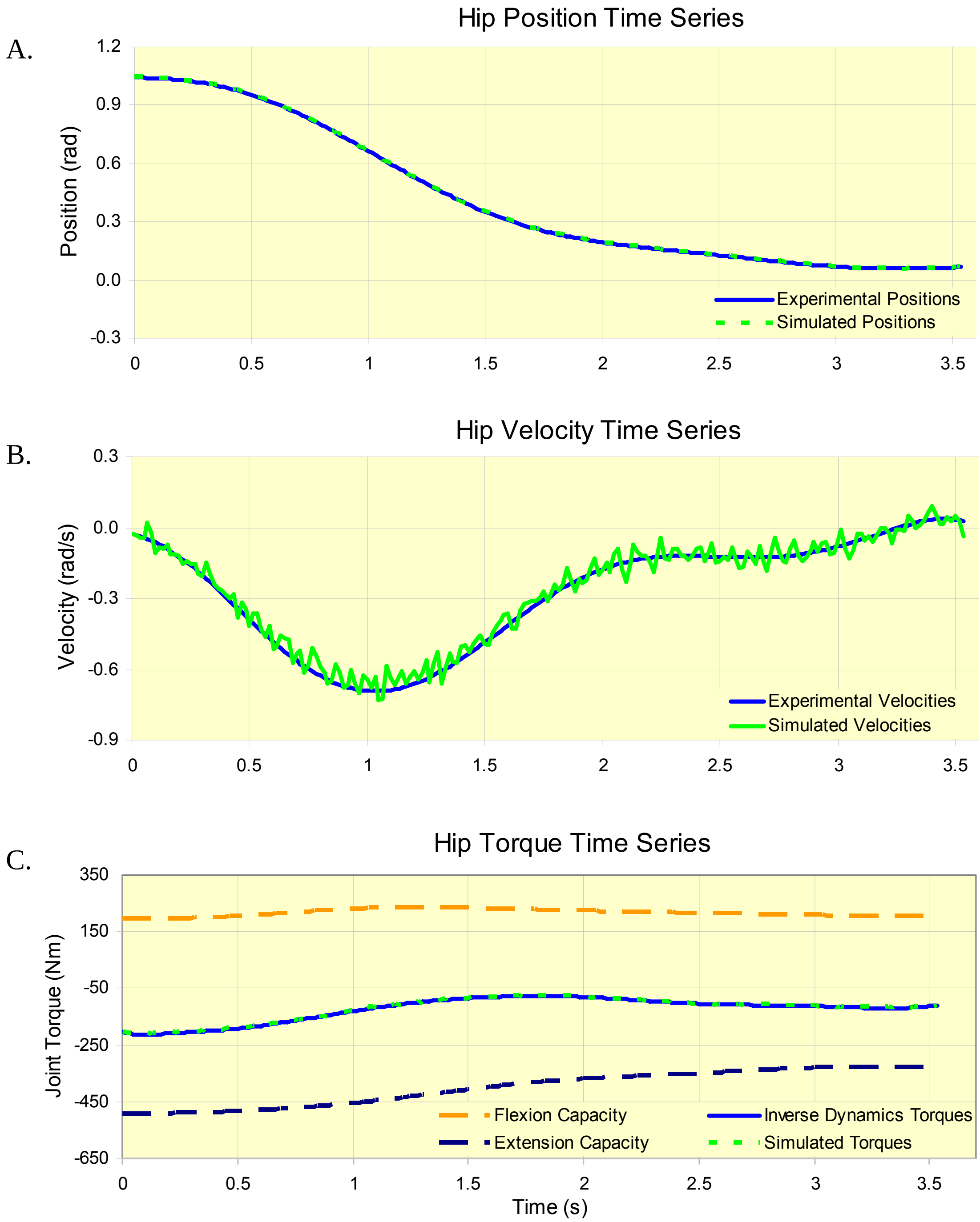


Figure C.3. Hip joint time series.

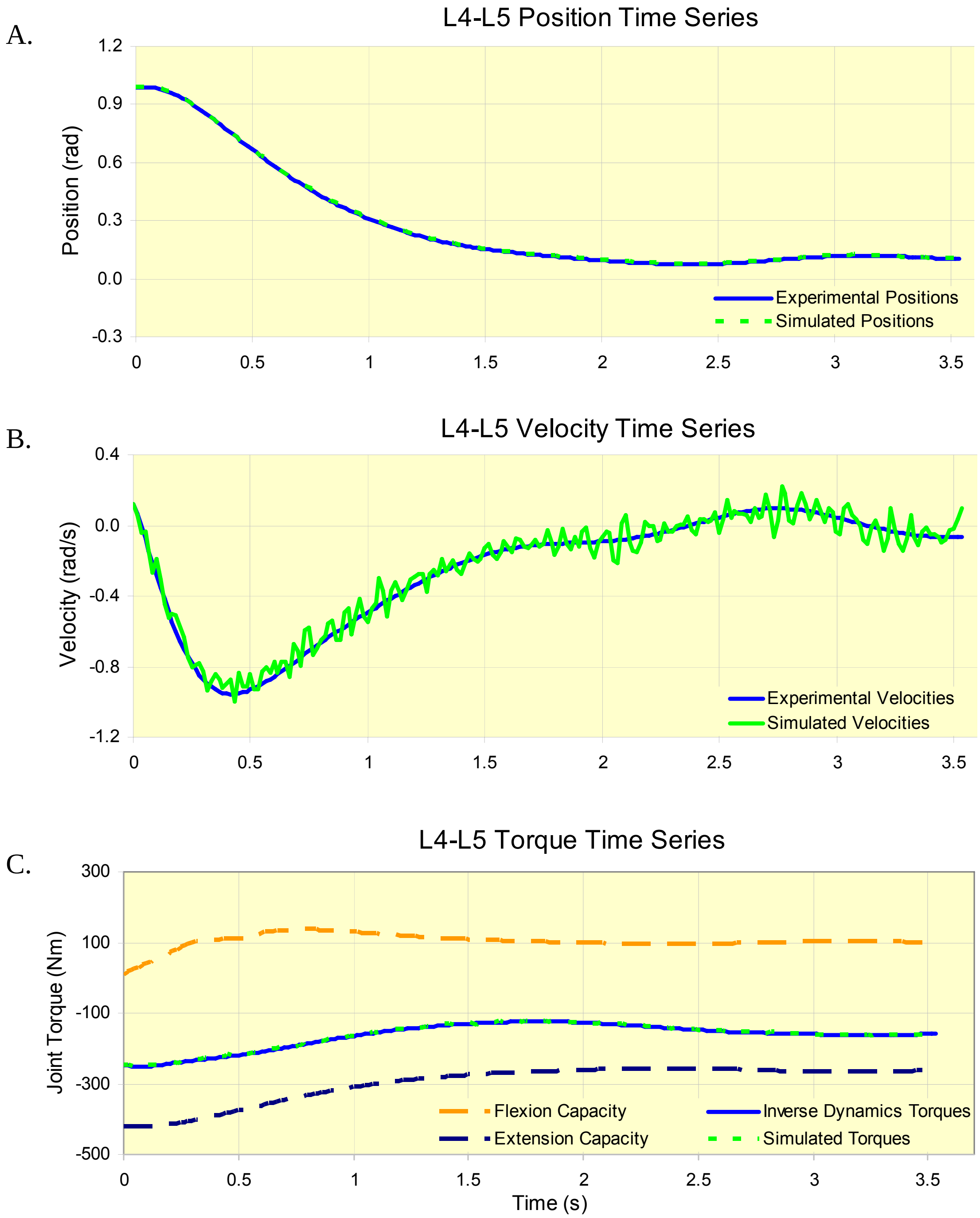


Figure C.4. L4-L5 joint time series.

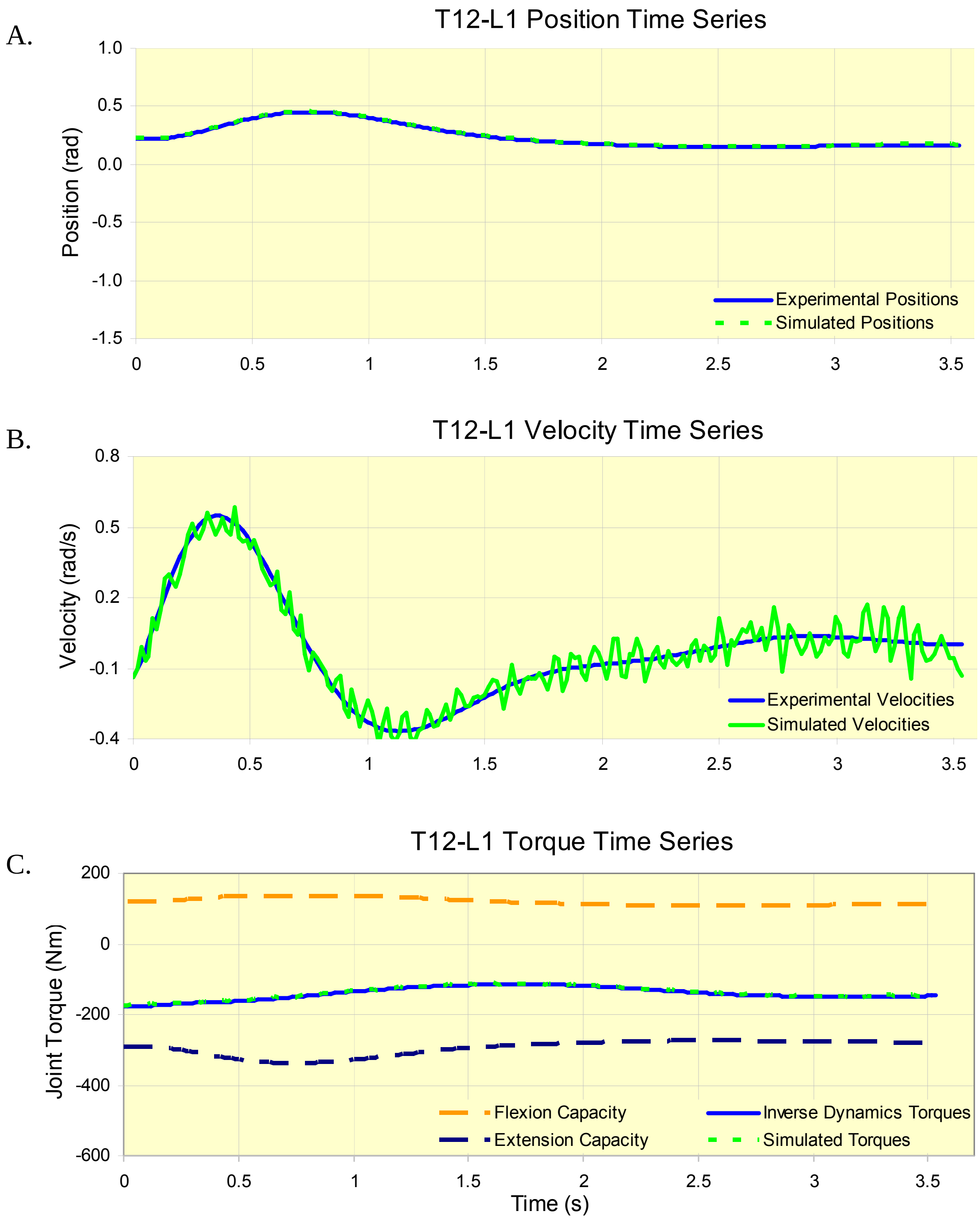


Figure C.5. T12-L1 joint time series.

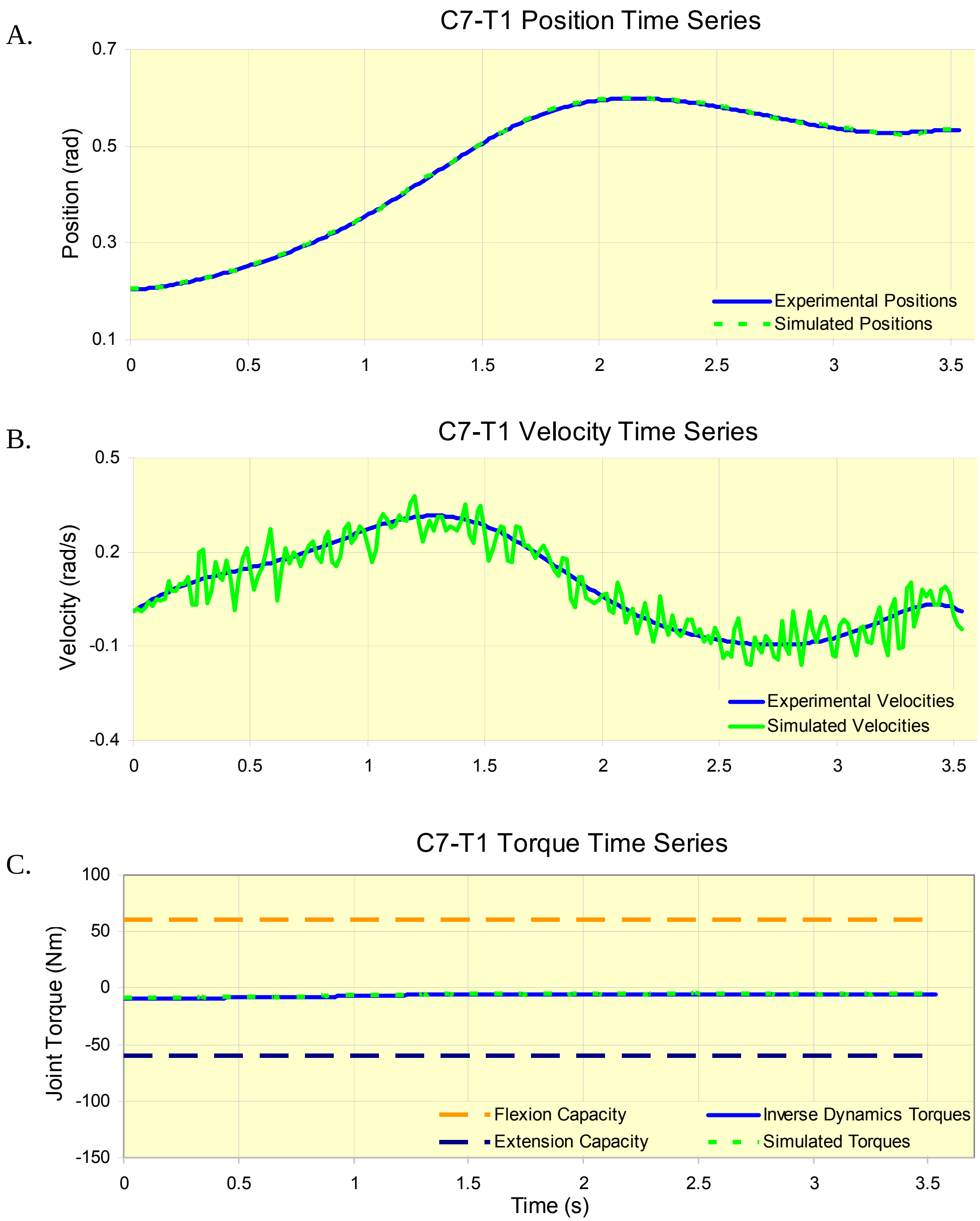
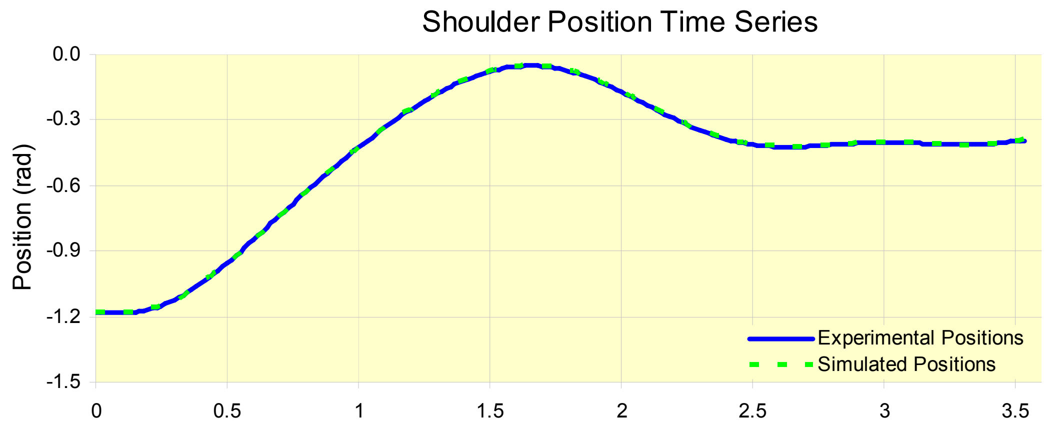
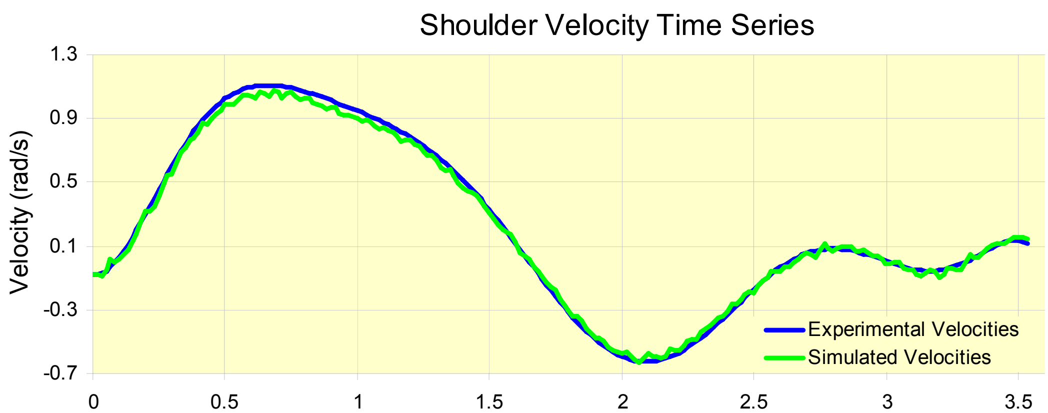


Figure C.6. C7-T1 joint time series.

A.



B.



C.

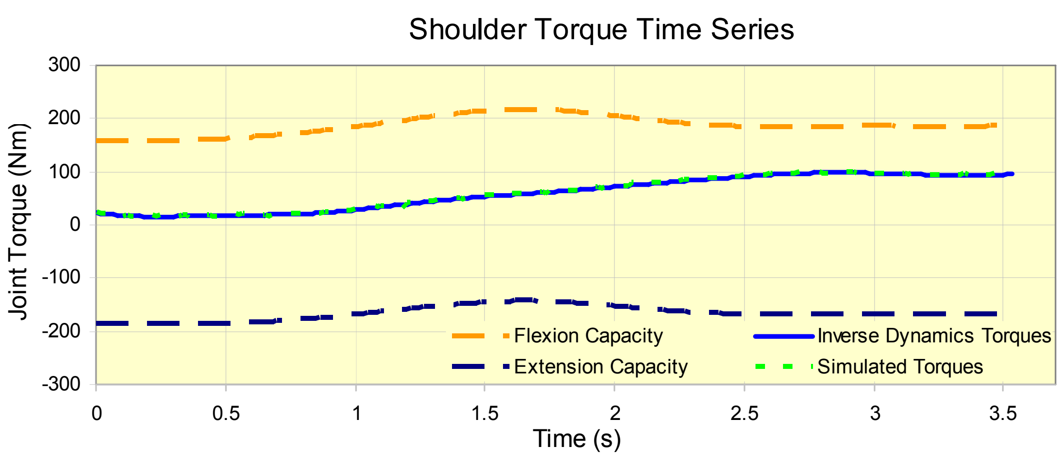


Figure C.7. Shoulder joint time series.

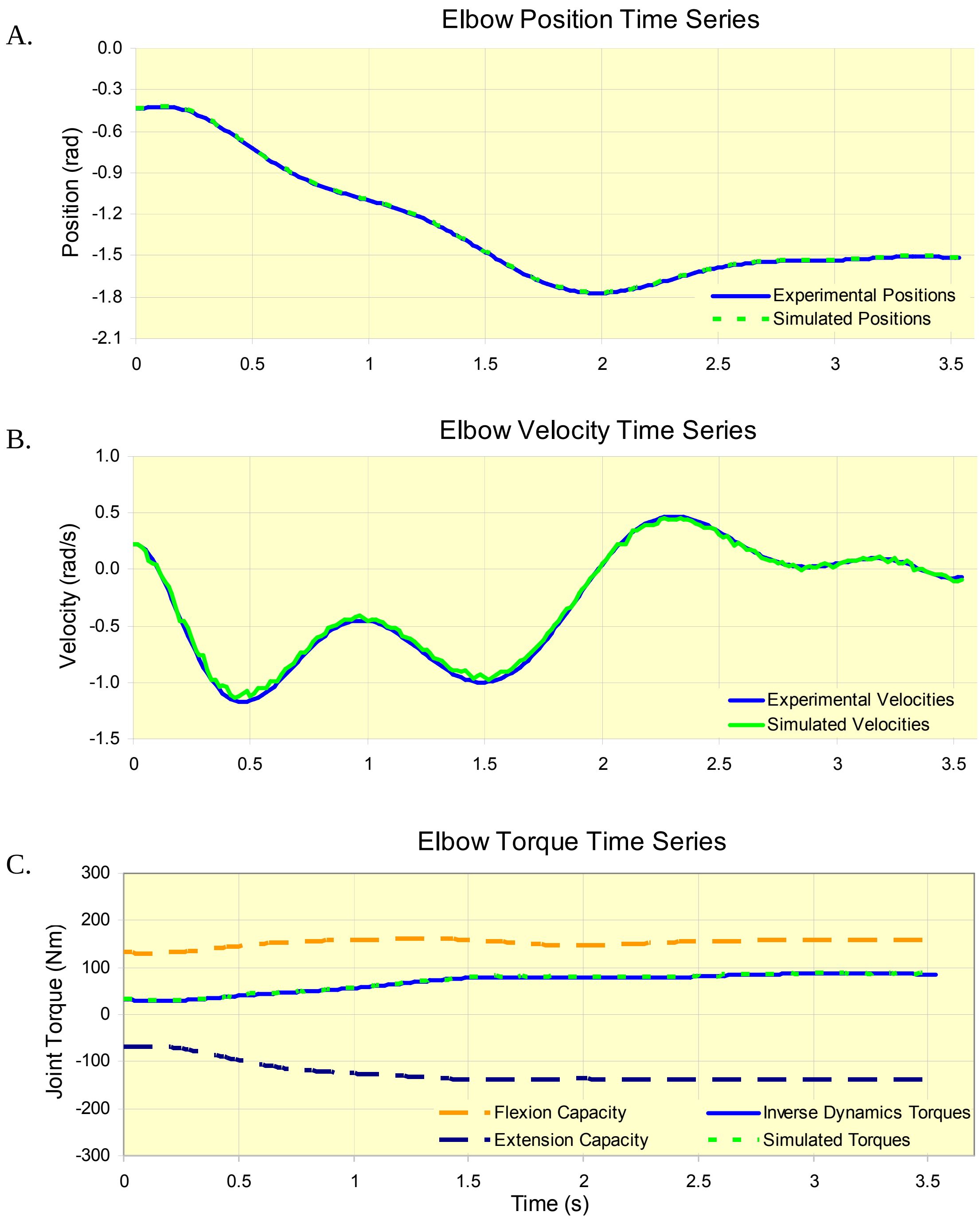


Figure C.8. Elbow joint time series.

APPENDIX D

Ground Reaction Force for each Scenario

Ground reaction forces computed by the model were validated by comparison to those measured from the force platform for the experimental trials. In Figure D.1, computed and measured foot COP are compared for (A) light, (B) medium, and (C) heavy experimental trials. Positive distances from the ankle indicate that the COP is toward the toe. Figure D.2 contains computed and measured vertical ground reaction force trajectories for (A) light, (B) medium, and (C) heavy experimental trials. Figure D.3 is identical to Figures D.2, except that the data is zoomed into closer range.

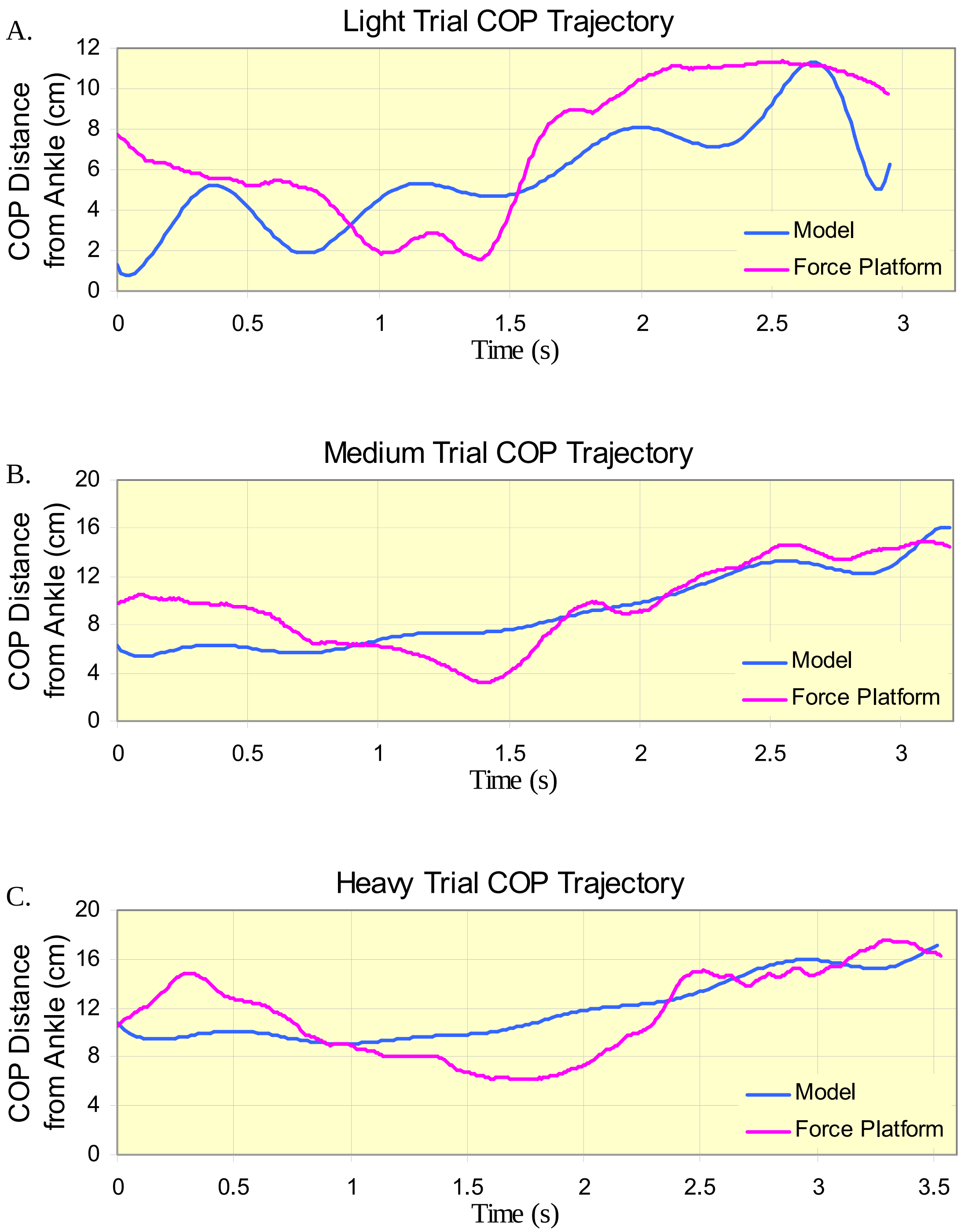


Figure D.1. COP trajectories of experimental trials.

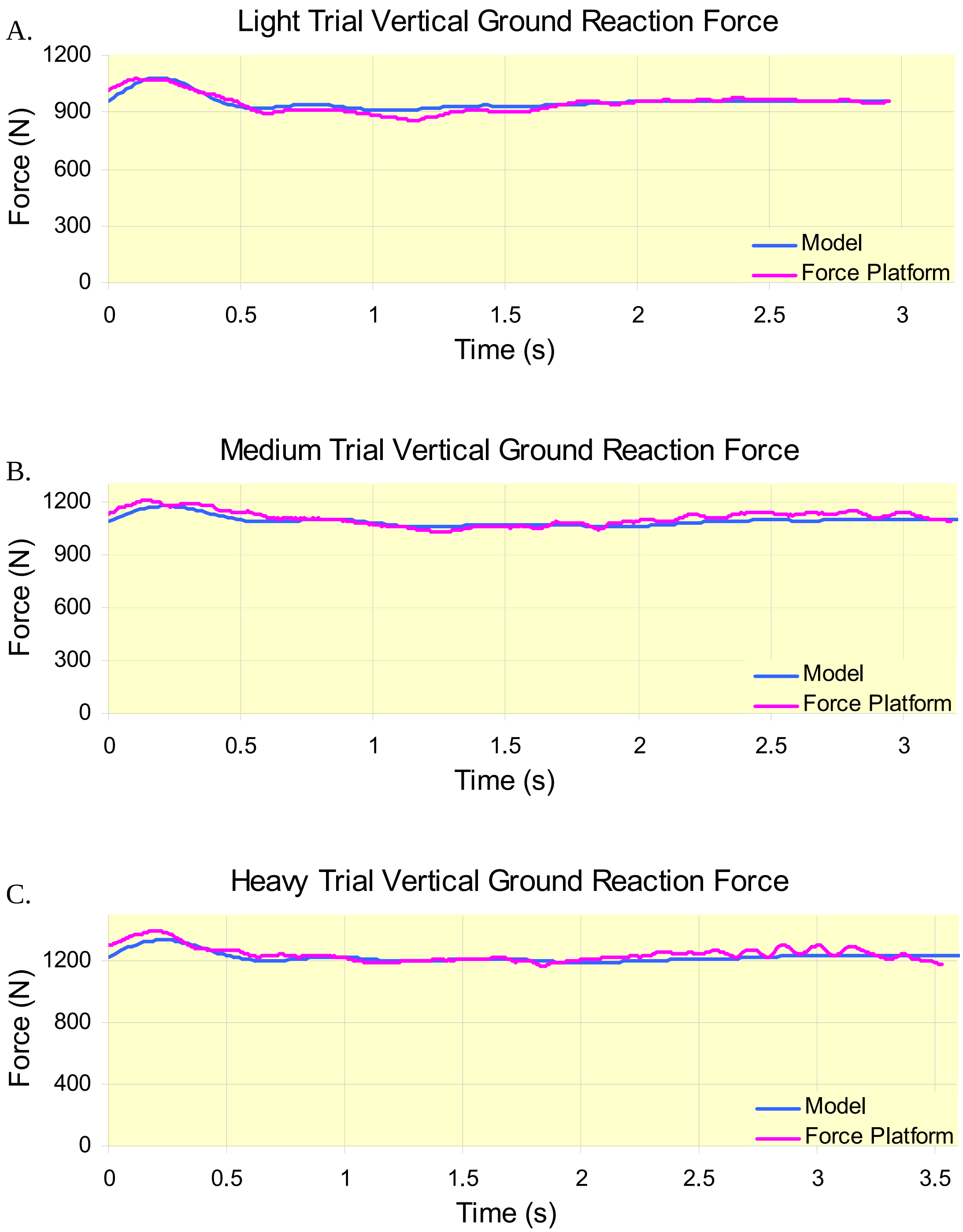


Figure D.2. Vertical ground reaction forces of experimental trials.

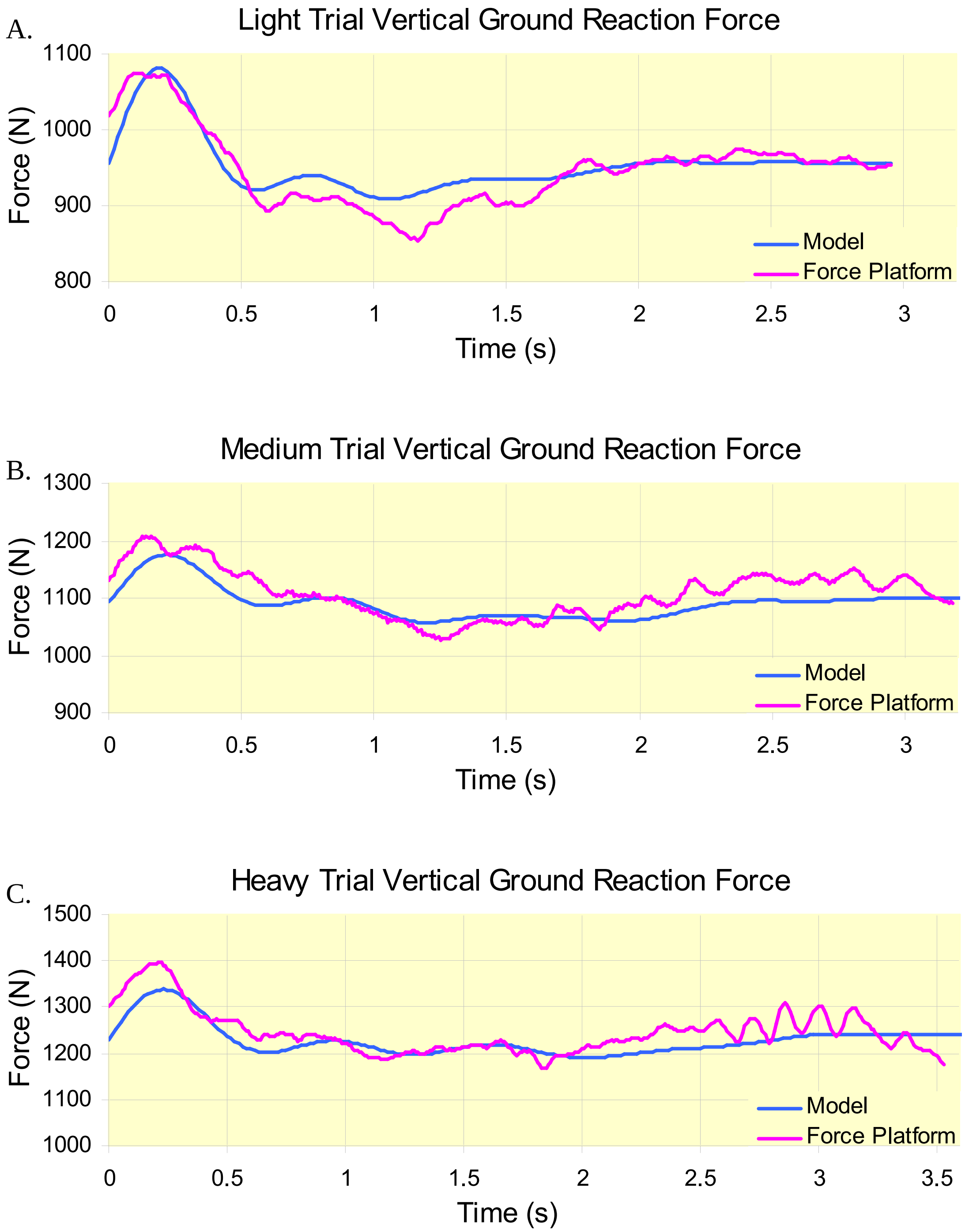


Figure D.3. Vertical ground reaction forces of experimental trials (zoomed in).

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