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I am submitting herewith a thesis written by Charles David Kohstall, Jr. entitled "The investigation of soft tissue vibrations and landing." I have examined the final paper copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Human Performance and Sport Studies.

Songning Zhang, Major Professor

We have read this thesis and recommend its acceptance:

Accepted for the Council:

Vice Provost and Dean of Graduate Studies

# The Investigation of Soft-tissue Vibrations and Landing.

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## Abstract

The purpose of this study was to examine the effects of muscle tone to an imparted vibration to the vastus lateralis during various mechanical demands (rest vs. back squat). Five healthy male college athletes and five healthy male college recreational athletes were subjected to imparted vibrations on the vastus lateralis muscle during rest (in the supine position) and during a back squat exercise at knee flexion angles of 30 and 35°, while three and five trials were collected respectively. Vibrations were measured from a tri-axial accelerometer securely attached to the skin overlying the vastus lateralis muscle. The model chosen for analysis was that for a damped sinusoidal oscillation. The frequency ($f$) and damping coefficient ($c$) were calculated directly from the acceleration signal. A 2x2 (knee angle x activation level) repeated measures analysis of variance (ANOVA) was used to evaluate frequencies and damping coefficients related to the vibrations. An increase in $f$ was associated with an increase in muscular contraction for the Z and Y-directions. No differences were shown between subject populations. The model chosen for analysis was unable to resolve any difference for the damping coefficients. It has been demonstrated that the frequencies and damping coefficients of the quadriceps increased with increased muscular force. The results of this study an inconclusive, but lend support that the frequencies increase with increasing muscular contraction.

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# Chapter I

## Introduction

Muscle tone indicates the conditioning of muscles and is an important aspect of athletic performance and injury prevention. During athletic training and competition, tremendous impact loading to the human body may occur in the landing part of sport activities. Prior research shows that the maximum impact force measured during a landing activity exceeded 14.4 body weights (BW) (23). The impact force must be attenuated throughout the body. Shoes, bones, cartilage and muscles are responsible for the dissipation of these impact forces, and muscles have been thought to have the greatest potential to achieve such a purpose (25).

Shock attenuation during landing has been studied extensively (4, 5, 20, 31). Research has typically focused on changes in impact attenuation due to different heights, techniques, and surface compliance. Dufek et al. (5) showed that during evaluation and prediction of impact force in landing, a flat-footed lander had a significantly greater second peak of ground reaction force (GRF) compared to a toe-heel lander. Research also suggests that impact forces increase as a result of increased landing height (5, 21, 31).

During landing, the body's potential energy converts into its kinetic form; the muscular work is responsible for the dissipation of the energy and impact force (4). The stiffness of the muscular system during landing is influenced by its rate of force and summation of twitches and is modified by touchdown geometry of the lower extremities (7). If the contact angle is held constant, then increases in landing height are responsible for increases in muscle activation. For further discussion, the stiffness of active muscle

will be referred to as "muscle tone" and its increase is a result of muscular tuning (26, 27).

Muscle activities associated with its tone can be recorded by three different non-invasive techniques. Electromyography (EMG) can be used to investigate the electrical activity of superficial muscles (15). The use of EMG is limited to the firing rate of a particular muscle at best. Mechanomyography (MMG) is the recording of harmonic vibrations of contracting muscles and has been used to describe increased tension and twitch summation in a contracting muscle (1). They showed an increase in MMG signals with an increase in percent of maximum voluntary contraction. However, results from some studies differ showing a plateau at 70% MVC or even a slight decrease in signal (6). Finally, mechanical vibrations of muscle can be studied by monitoring changes in the vibration frequencies and amplitudes of soft-tissue overlying a particular muscle using a three-dimensional accelerometer (26, 27).

Measuring soft-tissue vibrations can deliver a better understanding of coupling between the soft and rigid tissues of the body (26, 27). Muscles can be visually seen to twitch or oscillate during a contraction. Such vibrations are a result of muscular contractions; their frequencies are the result of the intensity at which the muscle contracts (1, 16, 17, 24). By measuring the changes in the coupling, changes in impact forces can be further evaluated.

**Problem Statement**

The purpose of this study was to examine the soft-tissue vibrations of the vastus lateralis muscle, and to examine the impact attenuation of landing. The frequencies and damping coefficients of soft-tissue vibrations associated with the vastus lateralis muscle

were studied using different activities (supine resting position, back squat). Impact attenuation was studied using landing heights (30, 45, and 60cm), landing techniques (soft and stiff). The soft-tissue vibrations and impact attenuation were studied using two different subject populations (Division I collegiate athletes of track and football teams and recreational athletes).

The quadriceps muscles are mainly involved in regulating the knee joint ROM during the landing phase. However, no studies to date have documented the frequency and damping characteristic of vibrations in the soft-tissue overlying the quadriceps muscles during landing activities or the back squat. This study investigated the frequencies and damping coefficients of the vastus lateralis muscle during rest and a back squat, while manipulating knee angle. Impact attenuation during landing was studied by manipulating landing height and landing technique. Past experiences and training background were also examined by using two subject groups: (1) Division I collegiate athletes from the track and football teams and (2) recreational athletes. Such information may provide better understanding of the mechanical properties of muscles directly involved in the attenuation of the impact forces. The results of this research may also provide better understanding of performance enhancement, injury mechanisms, and the mechanical differences between soft and stiff landings in the attenuation of impact.

**Hypotheses**

The following hypothesis were tested:

1) The frequency and damping coefficient of the vibration would increase with increased muscular contraction.

2) The frequency and damping coefficient of the vibration would increase with increased knee angle.

3) The frequency and damping coefficient of the vibration would be greater for subjects of increased muscle tone.

**Delimitations**

The study was conducted within the following delimitations:

1) Five Division-I intercollegiate male athletes (18-25 yrs) and five recreational athletes (18-25 yrs) selected from the student population at The University of Tennessee. Subjects had no histories of major lower extremity impairment.

2) Thirteen conditions of soft-tissue vibration tests: two conditions in a resting position with knee flexion angles of 30°, and 35°; five conditions in a back squat at knee angles of 30°, 35°, 45°, 65° and 85°; and six drop landing conditions from three different heights (30, 45, & 60cm) with soft and stiff landing techniques.

3) Biomechanical signals were collected and analyzed from a duration 100 ms prior to contact to the end of the landing phase for each landing trial, and for 10 seconds during each back squat trial.

4) Data was collected at 1200Hz from a force platform (AMTI), a tri-axial accelerometer (PCB, TSD 109F), and a electrogoniometer (Penny+Giles Biometrics Ltd). One digital video camera (JVC, GR-DVL 9800) was sampled at 120Hz.

**Limitations**

The study was limited by the following factors:

1) Subjects were limited to the student population at The University of Tennessee.

2) Possible errors from placement and digitizing of the reflective markers. Other errors such as perspective error and marker placement are acknowledged.

3) Inherent errors from the force platform, accelerometer, and/or digital video systems. Errors of force platform and high-speed video systems are always present but were considered acceptable within the specifications of the manufacturers. Errors caused by out of plane motion were controlled by confining the activity to the sagittal plane.

4) Potential errors due to the difference in sampling frequency of the force platform (1200Hz) and the digital video system (120Hz), and synchronization of the systems. Synchronization accuracy between the force and video systems was limited by the sampling rate of the slower system. The video system has a sampling error of ± 0.08 frames/second, resulting a maximum error of only 0.67 ms.

**Assumptions**

The following assumptions were made for this study:

1) The damping of impact due to muscle tone could be measured by the accelerometer.

2) Drop landings could be described and modeled mechanically.

3) Biomechanical measurements used were sufficient for analyzing effects of drop landings from different heights with different techniques on soft-tissue vibrations in the quadriceps muscle.

4) Biomechanical instruments used were accurate.

5) All subjects were free of injuries to their lower extremities at time of testing.

6) The performance of the subjects was symmetrical so that only the right side was assessed for kinematics, GRF, and acceleration.

# Chapter II

## Literature Review

### Landings Biomechanics

Dufek and Bates (5) investigated influences of landing height, distance and technique on the impact forces during landing activities. Three male subjects performed drop landings onto a force platform with simultaneous recordings of sagittal kinematics and ground reaction forces (GRF) in 27 conditions that were a combination of three landing heights (40, 60, and 100 cm), techniques (knee angle: stiff > 110°, slightly flexed 75 - 110°, and fully flexed < 75°) and distances (40, 70, and 100 cm). Significant results were found for both within subject and between subject analyses for the first (F1) and second (F2) GRF. The third subject was a flat-footed lander and skewed the results of the entire study. The within-subject comparisons showed no significant effects of height and technique for subjects one and two. The toe-heel landers showed lower F2 values than that of the flat footed subject.

Devita and Skelly (4) examined the effects of landing stiffness on attenuation of impact forces. Ground reaction force, joint kinematics, and joint kinetics of the lower extremity were compared between soft and stiff landings from a vertical fall of 59 cm. Eight female athletes performed 10 trials in each of two conditions: soft (knee flexion angle < 90°) and stiff (knee flexion angle > 90°) landing. The kinematics showed that subjects flexed more at each lower extremity joint in the soft landing. The mean time of landing phase (time from the contact to the maximum knee flexion) for the soft and stiff landings was 342 and 152 ms, respectively. The mean impact phase (time between initial contact and the moment when the vertical GRF reached either a minimum value or a

relatively stable value after F2) durations lasted 126 and 113 ms for soft and stiff landings. Knee extensor moments reached maximum values about 15ms after contact and ended about 25ms after contact. A hip flexor moment at contact accelerated the trunk forward and downward. Extensor moments were present at each joint for the remainder of the impact phase to reduce the subject's downward velocity.

Zhang et al. (31) investigated changes in energy absorption of lower extremity joints due to different landing heights and landing techniques. Nine healthy and active males performed step-off landings from three different heights (0.32, 0.62 & 1.03m) using three different techniques (soft-SFL; normal-NML; and stiff-STL). The results showed a general increase in peak GRF, peak joint moments, and power with increases in landing height and stiffness. The ROM for the ankle joint experienced small decreases from SFL to STL, with more significant decreases of ROM seen for the knee and hip across all three heights. Significant differences in F1, F2, and knee ROM were shown between the two landing techniques (SFL & STL) and the three drop heights (0.32, 0.62 & 1.03m). The eccentric work performed by the knee and hip joint extensors was decreased in stiff landings at each height.

**Muscle Stiffness and Damping and Body Stiffness**

Greene and McMahon (9) investigated the reflex stiffness of man's anti-gravity muscles during kneebends while carrying extra weights. At small knee flexion angles the reflex stiffness was independent of muscular force, but increased slightly for larger knee flexion angles. The results showed that flexing the knee resulted in one oscillation of the body to dampen motion whereas maintaining stiffness at the knee joint resulted in a few

oscillation of the system before restoration. The system of anti-gravity reflex muscles studied was characteristic of an underdamped system.

McMahon et al. (20) investigated the interaction between the effective vertical stiffness, gravity, impact velocity, and body mass using a simple mass-spring model. It was shown that vertical stiffness increases with running speed and that running with the knees flexed more than normal resulted in a decrease of mechanical shock transmitted to the skull (19, 20). The trade-off for decreased shock transmission was an increase in $O_2$ consumption by as much as 50%.

Lafortune et al. (15) investigated the role of interface and knee angle in impact cushioning and regulating initial leg stiffness using a pendulum apparatus. Three initial knee angles (0, 20, 40°) and three surfaced conditions (barefoot, and soft and hard EVA foams) were used during the testing of 21 subjects. The results showed the role of the knee joint angle in regulating initial leg stiffness was not significant and only a low level of correlation was shown between the knee flexion angles and improved cushioning. As the knee flexion angle increased the impact force decreased, but increased the shock traveling throughout the shank. The softer surfaces showed a reduction in both initial leg stiffness and magnitude of the impact experienced by the lower limb.

Conevey et al. (3) investigated the behavior of the leg during oscillation about the ankle joint. The response of the leg oscillation for 45 subjects showed non-linear behavior. The results showed that the gastrocnemius muscle and musclular tendinous unit (MTU) exhibited a decrease in stiffness with increasing deformation amplitude. This resulted in higher natural frequencies (shorter time periods) during successive oscillations.

Ferris and Farley (8) investigated the interaction of leg stiffness and surface stiffness during human hopping. It was shown that leg stiffness decreased as a result of increased surface stiffness. Leg stiffness can be regulated by a number of pathways and the control of movements may ultimately change due to environmental considerations. Farley et al. (7) investigated the mechanisms of leg stiffness for hopping on surfaces of different compliance. Mathematical models predict that leg stiffness should depend on touchdown leg geometry due to changes in the alignment of the ground reaction force vectors. It was shown that when subjects hopped on surfaces of different stiffness, leg stiffness increased as surface stiffness decreased. Further, leg stiffness was shown to be equal the ratio of the peak GRF to leg compression. It was also shown that leg stiffness was extremely sensitive to the stiffness of the ankle.

Arampatzis et al. (2) examined the effect of running velocity on leg stiffness and joint kinetics in human running. Thirteen subjects ran at velocities of 2.5, 3.5, 4.5, 5.5 and 6.5 m/s while GRF (Kistler) and kinematic data from two high-speed video cameras (120Hz) were sampled. The results indicated that the running velocity influenced the leg spring stiffness, effective vertical stiffness, and maximum vertical GRF. The modeled data were significantly lower than the data actually recorded. The modeled leg spring stiffness was not significantly different from the stiffness computed from the kinematic data. The spring stiffness at the ankle was not significantly different between the five running velocities.

Minetti, Ardigo, and Cotelli (21) investigated the use of leg muscles as a shock absorber in both theoretical predictions and experimental results of drop landing performance. The theoretical predictions were obtained using a phase-plane approach

and the experimental data were collected. The experimental groups consisted of sedentary (group one, n = 4) and elite skiers (group two, n = 36) with group one performing drop landings from three different heights (0.4, 0.71, & 1.10 m) and group two landing from a height of 0.75 m only. All subjects were instructed to stop downward movement as quickly as possible and land flat-footed. The result showed that the muscle force-velocity diagram for eccentric contraction speeds represented an increase in the negative power with increases in the drop height. The minimum duration of the landing phase depends on the sustainable peak force a subject can withstand while the maximum duration is determined by the maximum downward displacement available after touch down.

Nigg and Liu (22) examined the effect of muscle stiffness and damping on impact peaks in a simulation study for running. The simulation model of impact included a non-rigid mass ("wobbling" mass) that was mechanically coupled to rigid masses. The body was modeled as four masses. The shank, leg, and foot were represented as a single rigid mass (M1) and a wobbly mass (M2). Spring-damper units mechanically coupled with the upper and lower rigid masses, and the rigid and wobbling masses. Five spring constants (k) were also utilized to provide the necessary motion for impact force dissipation. A vertical touchdown velocity of 0.6m/s was used for all four masses; simulation conditions used combinations of two different midsole densities and three different spring constants. The results showed that deceleration of the lower rigid and wobbling masses during the impact phase was longer in the hard midsole than the soft shoe. The peak vertical GRF also increased with increasing stiffness of ankle and knee moments. However, changes

in damping coefficients had a greater influence on changes in the vertical impact force peak than changes in stiffness.

**Soft-tissue Vibrations**

Wakeling & Nigg (26, 27) investigated soft-tissue vibrations and characterized the damping coefficients of a vastus lateralis quadriceps muscle in relation to percent of maximum voluntary contraction (MVC). In the first experiment, subjects (7 men and 7 women) volunteered to participate in the experiment with soft tissue vibrations imparted on the muscle by a strike from a wooden mallet, 40mm proximal to the attachment site of an accelerometer that was placed at a position one third of the muscle length proximal to the distal end of the muscle belly (27). Measurements were recorded during isokinetic trials at the knee flexion angles of 20, 40, & 60° while the subjects were seated with the hip flexed at 85° and the thigh supported in a position parallel to the ground in a dynamometer. The vibration characteristics were analyzed in the time window from 50 to 250 ms after mallet strike.

The range of free vibration frequencies was from 9 to 32 Hz for the male and from 8 to 22 Hz for the female subject in trials with 0 - 100% MVC. A regression analysis showed a significant positive correlation between frequency (*f*) and damping coefficient (*c*) ($p<0.001$, $r^2=0.89$, normal to skin; $p=0.007$, $r^2=.38$, tangential to skin). Increases in the knee flexion angle caused increases in both *f* and *c*. The results showed the characteristics of an under-damped system for the free vibrations of the quadriceps; the amplitude of the free vibration reduced to 5% of its initial value within a duration of 40ms for the highest *c*, and within 400ms for the lowest *c*. Between the first and third cycles, the oscillation was shown to occur before the amplitude of the vibration was

damped to less than 5% of its initial value. The standard errors (SE) were greater for $c$ than for $f$, with the SE of $f$ never higher than 2.6 Hz and the SE of $c$ never greater than 9.6 $s^{-1}$. On average, the standard error of mean (SEM) was 0.02 and 0.07 for $f$ and $c$, respectively (27).

In the second study, another accelerometer was positioned 80mm proximal to the first to investigate the possible differences in positioning using similar setup as the first study (26). Significant differences were found between sites, but results showed no systematic bias of higher or lower values for $f$ and $c$. A Chi-square test showed that the proportions of higher and lower values between the distal and proximal accelerometer were not significantly different from random. The effects of muscle length and force production of soft tissue vibrations showed a general increase in both $f$ and $c$ with an increase in torque for all joint angles. The $f$ and $c$ values were smaller for lower joint torques than for higher torques (26).

During the isometric test, the quadriceps tissue showed mean f values of 8.2 to 41.3 $s^{-1}$ for $f$ and 8.4 to 61.5 $s^{-1}$ for $c$ when taken from the first and fourth quartile of the joint torque. The x direction vibration was most pronounced and showed $f$ and $c$ to increase the greatest during the highest joint torques when the muscle was most active. No significant effect occurred between either $f$ or $c$, and joint angle. Increases in the joint angular velocity produced significant increases in both $f$ and $c$ when data were analyzed using least-squares planar regression (26, 27). However, the data analysis was unable to resolve significant effects between the joint position (and thus the muscle length), and $f$ and $c$.

**Knee Joint Contact Angle during Landing**

In Zhang et al.'s (31)study on mechanisms of energy dissipation of lower extremity joints during landing, the mean value of minimum knee flexion angles was determined for each of nine landing conditions and the range of the minimum knee flexion angle was calculated using the following formula:

$$\text{Range of knee joint angle} = \text{mean} = \pm 9°$$

The range values were then used to monitor the actual landing trials during testing. The results showed that for the mean range of motion (ROM) for the knee joint from contact to max knee flexion for 32 cm landing height was 74.5°, 53.8°, and 27.9° for SFL, NML and STL respectively. For the drop height of 62 cm, the three techniques averaged 76.1°, 58.6°, and 35.0°. Finally, the 103 cm drop height showed means of 80.6°, 63.8°, and 47.0°, for SFL, NML and STL respectively. In addition, the time to the minimum position of center of gravity (COG) was less than 200 ms for stiff landings and was around or greater than 300 ms for soft landings (31).

Yu (30) provided data of the knee contact angle in landing. This study used four heights of 30, 45, 60 and 75 cm and two surfaces (shoes or no shoes). The knee contact angles were 29.6 ± 8.0°, 32.3 ± 6.9°, 31.6 ± 7.9°, and 33.0 ± 6.6° for the four previously mentioned heights with shoes. When the group means for the toe-heel versus the flat-footed landers were compared, no significant differences were found. This finding and the other landing studies conducted in the Biomechanics Lab at UTK (personal communications) indicate that 30° of knee flexion is a consistent and reliable assessment of knee contact angle for landings at moderate heights.

**Stiffness of Muscle**

Rack and Westbury (24) investigated the short-range stiffness of active mammalian muscle and its mechanical properties. Experiments were carried out on the soleus muscle of eleven cats. Length and tension were recorded on magnetic tape by frequency modulations. A decrease in the mean tension was seen when high frequencies and large amplitudes of movement were used, and it was greater when the rate of stimulation of the muscle was low. However, once the initial fall in tension was over, the response to succeeding cycles of movement remained fairly constant. Further, the fall in mean tension during the course of an alternating movement was accompanied by a reduction in the extent of the tension fluctuation even though the amplitude of movement remained the same. At all stimulation rates tested, the data showed a rising trend for muscle tension during lengthening and a falling trend during shortening. A higher resistance to faster movements was shown, and the overall tension changes in fast alternating movements were always greater than in the slower trials. The results showed that resistance of the muscle to movements was dependent on the intensity with which it was activated and that the short-range stiffness varied proportionally to the mean muscle tension.

Accordingly, the relationship between the amplitude of movement and the peak-to-peak tension change may be used as a measure of the effectiveness of the muscle at resisting the whole movement. Further, the amount of work required to drive the muscle through a complete cycle of movement gives a measure of the ability of the muscles to absorb and dissipate mechanical energy. The experimental data showed that for movements of short duration the tension changes were largely determined by the short-

range stiffness of the muscle since it was never stretched past this. As the amplitude of movement increased, the associated tension changes fell further below the level measured for small movements. During small movements at moderate to fast velocities, the muscle behavior was like that of a stiff spring, which resisted extension with a steeply rising force, but during shortening yielded up most of energy that was expended on it during lengthening. On the other hand, during large movements the behavior of the muscle was similar to a compliant spring that was heavily damped. The data showed a relatively smaller rise in force during lengthening and an appreciable part of the work that went into the lengthening movement was dissipated by the muscle and did not re-emerge during shortening. Finally, the short-range stiffness that occurred in the early part of a movement (pre-activation) could reflect the combined stiffness of a large number of actin myosin cross-bridge links activating simultaneously (24).

Arampatzis, Bruggemann, and Klapsing (2) examined the effect of verbal instructions given to subjects on the control of lower extremity stiffness and the effects of leg stiffness on mechanical energetic changes during drop jumps on a compliant surface. The instruction of "jump high faster" resulted in a higher level of muscle activation that caused increased leg stiffness. Increased leg stiffness then resulted in higher maximal GRF, decreased vertical displacement of COG, and shorter contact times.

**Mechanomyography**

Muscle vibrations resulting from coupling of series parallel elements has been studied by use of mechanomyography (MMG) that employs piezoelectric sensors and microphones (6, 14, 16-18, 29).

Ebersole et al. (6) investigated MMG and EMG responses of the superficial quadriceps femoris muscles using isometric contractions at knee flexion angles of 25, 50, and 75° while performing at 25, 50, 75, and 100% MVC. Results showed that the MMG amplitude increased for each muscle group up to 100% MVC at knee flexion angles of 25 and 50° only. A plateau of the MMG signal was observed from 75 to 100% MVC at the knee flexion angle of 75°. The authors hypothesized that the varied results between subjects at the different knee angles were the result of differences in muscle stiffness, intramuscular fluid pressure, and/or motor unit firing frequency.

Madeleine et al. (18) investigated relationships between mechanomyography and electromyography, and force during concentric, isometric, and eccentric contractions from the first doral interosseus muscle of 13 subjects. A non-linear relationship was found between the recorded EMG and MMG signals, which was dependent on the type of muscular contractions, contraction levels, and angular velocities.

Kouzaki et al. (14) investigated mechanical activities of the quadriceps muscle during fatigue by repeated maximal voluntary contraction. Seven subjects performed baseline tests at 10% increments from 40 to 90% MVC for 3-second intervals followed by 1-minute of rest. Then subjects performed 50 MVCs for 3-second with 3-second rest to achieve fatigue in the quadriceps muscle group. The mean MVC force fell 49.5% at the end of the fatigue test and a non-uniform decline in integrated MMG was found. The decrease in recorded MMG signal was most pronounced in the rectus femoris, followed by the vastus medialis, and least in the vastus lateralis.

Weir et al. (29) investigated MMG and EMG responses during fatigue in humans during isometric dorsiflexion contractions from 40° of plantarflexion and 5° of dorsiflexion. The investigators only used 50% MVC to fatigue the tibialis anterior muscle. MMG and EMG amplitudes increased with fatigue while the frequency decreased at both muscle lengths. It was concluded that larger increases in motor unit recruitment occurred with time during the fatigue test.

# Chapter III

## Research Methods

### Experimental Methods

The purpose of this study was to examine the soft-tissue vibrations of the vastus lateralis muscle, and to examine the impact attenuation of landing. Amplitudes, frequencies and damping coefficients of the muscle, and the biomechanical response were examined.

#### Subjects

Five Division-I collegiate male athletes (Age: 21.0 ± 2.2 yrs, Body mass: 82.2 ± 8.9 kg, Height: 180.5 ± 7.6 cm) and five recreational athletes (Age: 23.0 ± 1.2 yrs, Body mass: 83.5 ± 9.3 kg, Height: 181.4 ± 2.9cm) volunteered to participate in the study. The collegiate athletes were participants of the track and/or football team and had a self purported 40 yard dash time of less than or equal to 4.6 seconds and vertical jump of greater than or equal to 35 inches. Recreational subjects were from the student population on The University of Tennessee campus. The subjects had no histories of major lower extremity impairments and participated in recreational activities on a regular basis (2-3 times per week). Each subject signed an Informed Consent Form consistent with the human subject policy of the Institutional Review Board of The University of Tennessee prior to the experimental sessions.

#### Instrumentation

All tests were conducted in the Biomechanics/Sports Medicine Lab, Room 135, HPER Building at The University of Tennessee.

**Kinematics**

One high-speed video camera (GR-DVL9800, JVS) was used to obtain right-side sagittal kinematic data of the subjects at 120Hz. A reference frame (width = 140.97 cm, height = 186.69cm) was used to obtain scaling factors for the conversion of anatomical coordinates of reflective markers from an image reference to a lab reference. Reflective markers were placed on the right side of the body at the mid-rib, hip, knee, ankle, and head of the fifth metatarsal (Figure H-1). The Ariel Performance Analysis System (APAS, Ariel Dynamics, Inc.) was used to capture and digitize video images. The digitized coordinates were then analyzed using customized software to determine the time-history and discrete events of linear and angular positions, velocities, and accelerations.

**Force Platform**

A force platform (OR6-7, AMTI) flush with the surrounding floor was used to collect ground reaction forces (GRF) and moments during the test. The GRF data measured Fx, Fy, and Fz, representing medial-lateral, anterior-posterior, and vertical forces respectively. The moments applied about the Fx, Fy, and Fz axes are Mx, My, and Mz, respectively. Signals from the force platform were sampled for 1.0 sec at a frequency of 1200 Hz and amplified before storage on the APAS computer through an A/D converter.

**Accelerometer**

One tri-axial accelerometer (TSD 109F, PCB) was used to measure soft-tissue vibrations of the vastus lateralis muscle at a position 40cm proximal to the distal end of

the muscle belly. Signals from the accelerometer were also sampled at a frequency of 1200 Hz using APAS. The accelerometer has a mass of 17g, a range of 50g, and a resolution of .005g. The accelerometer was securely fastened to the subject by use of double sided adhesive tape and elastic wrap.

**Electrogoniometer**

An electrogoniometer (Penny+Giles, Biometric Ltd.) was used to monitor knee flexion angles during the test to ensure proper landing stiffness. The electrogoniometer was sampled at 1200 Hz using APAS simultaneously with other analog signals. The signal collected from the electrogoniometer was converted and displayed on the computer screen after each trial. The range of maximum knee angle for each test condition was determined at the beginning of each condition. Three trials of landing were performed to obtain the average of the maximum knee flexion angle. The range of the angle was then determined with this equation: $\alpha_{range} = \theta \pm 9°$ (31). The maximum knee angle for each trial during testing was visually inspected to assure if the trail was within the range in each condition.

**Synchronization**

The signal from the force platform, accelerometers, and video were simultaneously recorded during the experiment using a customized trigger system with a light emitting diode (LED).

**Shoes**

All subjects wore identical lab shoes (Adidas) provided by the Biomechanics/Sports Medicine Lab.

**Experimental Protocol**

Prior to their participation in this study, subjects were informed of the purpose and the procedures of the study by the principal investigator (PI). The test was conducted on two separate days with a minimum of one-day rest and no more than four days in between testing. On the test day, subjects were further informed about the purpose, the number of conditions, the number of repetitions, and performance requirements of the study. The first session lasted 30 minutes and the second session lasted approximately 1 hour, including practice and getting familiar with the testing protocol.

In the first test session anthropometric measurements were collected. The weight of the subjects, and the girth and length of their lower extremity segments were measured and recorded. Weights were measured on the force platform, lengths were measured with a standard meter stick, and girths were measured with a flexible tape measure. Each measurement was repeated three times and the average was used for subsequent data analyses. Subjects performed 10 warm-up squats (with a weight equal to their BW) to ensure proper technique and warm-up. Then subjects were asked to perform back squats with a spotter(s) present. Muscle vibrations were recorded using the accelerometer after a strike by a mallet 40mm proximal to the attachment site of the accelerometer during the squats. Subjects performed 1 back squat at five different knee flexion angles of 30°, 35°, 45°, 65° and 85° for a total of five trials. Subjects were asked to hold at the required knee flexion angle in each squat for 10 seconds while a muscle vibration was imparted on the vastus lateralis muscle at the above-mentioned position by the PI. In addition, three trials of muscle vibrations were recorded with subjects lying in a supine position and the right knee supported with contoured foam and flexed at 30° and 35° while the muscle vibration

was initiated at the same position using the same method. The knee flexion angle was monitored by qualified lab personnel using a hand-held goniometer. The peak acceleration was monitored for consistent mallet strikes (26, 27).

In the second test session, drop landings were performed. The subject was asked to return for the second session with a minimum of one-day rest, but no more than four days after the first session. At the beginning of each drop landing condition, the subject performed three drop landing trials with the maximum knee joint angle obtained from the electrogoniometer to determine the range of the maximum knee flexion. The range was used to monitor the landing technique during the actual testing. The subjects performed five drop landings in each of six test conditions with the right foot on the force platform and the left foot on the adjacent floor flush with the force platform, for a total of 30 landing trials. The six test conditions are comprised of combinations of landings from 30, 45, and 60 cm with soft and stiff landing techniques. Subjects were instructed to land "as quickly as possible" and "with as little knee flexion as possible" for the stiff landing and "as softly as possible" for the soft landing. The order of drop heights and landing techniques was randomized. During the test session, the maximum knee flexion was monitored for each trial. Any trial where the maximum knee flexion angle exceeded the predetermined range was repeated. The diagram of experimental setup is shown in Figure G-2.

## Data Reduction

### Vibrations

The dominant frequency and damping coefficient at that frequency were calculated for the vibration signals from each axis of the accelerometer. These coefficients were determined using the following equation

$$S = Ae^{-ct} \sin(2\pi ft + \varphi)$$

where

$S$: the measured signal

$A$: the amplitude of the vibration

$t$: time

$c$: the damping coefficient

$f$: the damped natural frequency, and

$\varphi$: a phase coefficient.

The model chosen for analysis was that for a damped sinusoidal oscillation (26, 27). The frequency ($f$) and damping coefficient ($c$) were calculated directly from the acceleration signal. The advantage to this approach, according to Wakeling et. al. (27), is that it does not require an assumption of initial velocity or position that would otherwise be invoked if an integration of the acceleration were performed. In this system, frequency and damping characteristics in the soft tissues are purported to be the interaction either between the mallet strike and the muscle tissue during the imparted free-vibration tests or between the impact loading and the muscle tissue during the landing trials. After cessation of the impulse during the free-vibration trials the signals were completely due to the biomechanical properties of the tissue (26, 27). The largest

vibration, incorporating the peak acceleration, was chosen for frequency analysis. The damping coefficient was determined for the time window from the start of the largest oscillation to the end of the vibration signal.

The oscillation frequency and damping coefficient were compared under several scenarios. Wakeling et al. (27) reported frequencies and damping coefficients for the quadriceps muscle under 0, 50, and 100% max voluntary contraction (MVC) during isometric and isokinetic contractions at 40° of knee flexion. The calculated frequencies and damping coefficients from the current study in the free-vibration conditions were compared to those previously reported.

**Kinematic Data**

The kinematic variables were processed from the digital video in the following steps. A total of 121 frames of video images were captured for each trial with 20 frames prior to and 100 frames after foot contact with the force platform, and stored on the APAS computer. The reflective markers were then digitized using the APAS digitize module. The reference frame was also digitized to obtain scale factors to convert the coordinates of digitized reflective markers from a screen reference system to a lab reference system. The digitized coordinates were then decoded, smoothed, and reconstructed using a customized computer program. A Shannon algorithm was used to reconstruct the video signal from 120 Hz to 240 Hz (10). The digitized coordinates were smoothed using an algorithm to obtain optimal cutoff frequencies for the x and y coordinates of each reflective marker (13). The decoded time-history file was imported into the second customized program to compute linear and angular kinematic variables and determine corresponding critical events. These variables included contact angle and

velocity, maximum and time to maximum angle and velocity, and minimum and time to minimum position for the hip, knee and ankle joints.

**Kinetic Data**

The data collected from the force platform and the accelerometer were analyzed first by decoding the analog file to obtain the time-history of GRF and acceleration data. Then the decoded GRF files were imported into a custom program to compute and obtain selected variables. The GRF variables included the first (F1) and second (F2) maximum vertical GRF and the times (T1 and T2) at which they occurred, the loading rate of F1 and F2, and the impulse from contact to 100ms. The decoded accelerometer files for the squat conditions were truncated in a custom program to obtain each of the five trials. Then the accelerometer files were imported into a custom program to compute the peak amplitude, dominant frequency, and damping coefficients of the vastus lateralis from the imparted impact.

**Statistical Analysis**

Selected GRF and kinematic variables were analyzed in a 2x3 (technique x height) repeated measures analysis of variance. The selected accelerometer variables were analyzed in a 2x2 (knee angle x contraction level) and a 2x2x3 (knee angle x contraction level x channel) repeated measures of analysis of variance. Post-hoc comparisons were analyzed by Least Significant Difference (LSD). The statistical procedures were computed using SPSS statistics program (version 11.0) with the significant level set at $p<0.05$. The individual subject results of vertical ground reaction force, kinematics, and accelerations are provided in Appendix D, E, and F, respectively.

## Chapter IV

## Results

The purpose of this study was to examine the soft-tissue vibrations of the vastus lateralis muscle, and to examine the impact attenuation of landing.

For the X-axis (medio-lateral to the long axis of the thigh) accelerometer signal, the results of the ANOVA indicated no significant main effects. However, the post-hoc comparison did reveal a significant increase in the amplitude at 35° of knee flexion between the relaxed and squat conditions and a significant increase in AmpX from 30° to 35° knee flexion during squat conditions (Table 1, see Appendix A for variable definitions).

The results of the ANOVA for the Y-direction (anterior-posterior to the long axis of the thigh), revealed a significant main effect of muscular contraction for the amplitude.

**Table 1. Means and Standard Deviations of Selected Accelerometer Variables**

| Condition | FreqX | CX | AmpX | FreqY | CY | AmpY | FreqZ | CZ | AmpZ |
|---|---|---|---|---|---|---|---|---|---|
| 30° Relax | 69.095 | 105.082 | 2.809$^{A}$ | 53.995$^{\delta}$ | 50.665 | 8.781$^{\delta}$ | 42.562$^{\delta C}$ | 52.691 | 7.499$^{\delta C}$ |
| | (53.543) | (171.563) | (1.083) | (13.520) | (44.396) | (4.375) | (24.983) | (68.507) | (3.589) |
| 35° Relax | 61.780 | 30.611 | 3.034$^{\delta A}$ | 65.418 | 68.808 | 8.471$^{\delta}$ | 41.306$^{\delta C}$ | 50.178 | 8.093$^{C}$ |
| | (37.90) | (63.044) | (1.152) | (35.266) | (105.252) | (3.993) | (26.241) | (59.461) | (2.792) |
| 30° Squat | 74.265 | 86.728 | 3.590$^{\lambda A}$ | 65.402$^{\lambda}$ | 81.280 | 15.278$^{B}$ | 61.959 | 78.756 | 5.246 |
| | (40.801) | (314.344) | (1.906) | (11.451) | (102.056) | (7.333) | (19.774) | (97.719) | (3.969) |
| 35° Squat | 75.239 | 69.818 | 4.443$^{A}$ | 75.180 | 80.482 | 15.710$^{B}$ | 70.647 | 41.042 | 5.890 |
| | (27.945) | (123.640) | (2.070) | (18.723) | (84.873) | (9.697) | (23.218) | (58.893) | (5.745) |

Note:
$^{\lambda}$. Significant difference between knee angles for the contraction level
$^{\delta}$. Significant difference between contraction levels for the knee angle
$^{A}$. Significant difference between channel X and Y for the condition
$^{B}$. Significant difference between channel Y and Z for the condition
$^{C}$. Significant difference between channel Z and X for the condition
Frequency (Freq) is in Hz.
Damping Coefficient (C) is in $Hz^{-1}$
Amplitude (Amp) is in g.
The definitions of variables are in Appendix A.

Post-hoc comparisons revealed a significant increase for AmpY at both 30° and 35° knee flexion from the relaxed to squat conditions (Table 1). Post-hoc comparisons also indicated a significant increase for FreqY at 30° knee flexion from the relaxed to squat condition. A significant increase also occurred for FreqY from 30° to 35° knee flexion during squat conditions.

The results of the ANOVA for the Z-direction (parallel to the long axis of the thigh) indicated a significant main effect of the muscle contraction level for FreqZ. Post-hoc comparison revealed that the frequency of the free vibrations significantly increased from the relaxed to squat conditions (Table 1). The results also showed a significant decrease in AmpZ from the relaxed to squat conditions at 30°.

Results of the ANOVA, for the comparisons among three channels (X, Y, & Z-direction), indicated a significant main effect of amplitude. The post-hoc comparison showed a significant increase in amplitude from the X to Y-channel in all conditions (Table 1). Further, the Y-amplitude was significantly greater than Z in both squat conditions, and the Z-amplitude was significantly greater than X in both relaxed conditions. The results revealed the Z-channel frequency was significantly less than the X-channel for both relaxed conditions.

The results of the ANOVA, for Vertical Ground Reaction Forces, indicated a significant main effect of technique and height for F1, F2, LRF1, LRF2, and Impulse (see Appendix A for definitions). The post-hoc comparison showed a significant increase for F1, F2, LRF1, LRF2, and Impulse from SFL to STL at each height (Table 2). The results showed that F1, F2, LRF1, and Impulse significantly increased as the height increased for

**Table 2. Means and Standard Deviation of Vertical GRF Variables**

| Condition | F1 | F2 | LRF1 | LRF2 | Impulse |
|---|---|---|---|---|---|
| 30cm, SFL | $9.121^{\alpha\dagger}$ | $28.017^{\alpha\dagger}$ | $770.748^{\alpha\dagger}$ | $617.021^{\alpha\dagger}$ | $1.414^{\alpha\dagger}$ |
| | (2.496) | (9.592) | (461.361) | (724.544) | (0.256) |
| 30cm, STL | $18.099^{\alpha}$ | $51.853^{\alpha}$ | $1483.189^{\alpha}$ | $1088.981^{\alpha}$ | $2.688^{\alpha}$ |
| | (5.651) | (10.304) | (887.482) | (925.474) | (0.477) |
| 45cm, SFL | $17.395^{\beta\dagger}$ | $40.563^{\beta\dagger}$ | $1595.523^{\beta\dagger}$ | $1027.797^{\dagger}$ | $1.979^{\beta\dagger}$ |
| | (4.151) | (14.838) | (642.874) | (963.818) | (0.239) |
| 45cm, STL | $27.169^{\beta}$ | $64.828^{\beta}$ | $2265.417^{\beta}$ | 1548.635 | $3.159^{\beta}$ |
| | (9.944) | (14.389) | (865.169) | (1092.991) | (0.543) |
| 60cm, SFL | $27.945^{\gamma\dagger}$ | $50.316^{\gamma\dagger}$ | $2453.598^{\gamma\dagger}$ | $1287.382^{\gamma\dagger}$ | $2.392^{\gamma\dagger}$ |
| | (8.731) | (15.599) | (1019.418) | (925.390) | (0.259) |
| 60cm, STL | $36.392^{\gamma}$ | $73.293^{\gamma}$ | $3093.133^{\gamma}$ | $1901.907^{\gamma}$ | $3.418^{\gamma}$ |
| | (13.831) | (16.325) | (1496.567) | (1070.230) | (0.466) |

Note:
α. Significant difference between 30 and 45cm for the same landing technique.
β. Significant difference between 45 and 60cm for the same landing technique.
γ. Significant difference between 30 and 60cm for the same landing technique.
†. Significant difference between SFL and STL at the same height. Force unit is in N/kg and time is in second. Loading rate unit is in N/kg/s. Impulse unit is in (N/kg)s
The definitions of variables are in Appendix A.

the same landing technique. The LRF2 significantly increased from 30cm to both 45 and 60cm for the same landing technique.

The results of the ANOVA, for the ankle joint, indicated a significant main effect of height for ContVel and MaxVel. The post-hoc comparison showed a significantly greater ContVel and MaxVel at each height for SFL (Table 2). ContVel showed a significant increase from 30cm to both 45 and 60cm in STL; MaxVel showed a significant increase from 30 to 45 cm in STL. A significant increase was also shown for MaxVel from SFL to STL at 45cm.

The results of the ANOVA, for the knee joint, showed a significant main effect of technique for ContAng, ROM, and ContVel and a significant main effect of height for ContVel and MaxVel.

**Table 3. Means and Standard Deviations of Selected Kinematic Variables**

| Joint | Condition | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| Ankle | 30cm, SFL | -16.206 | 36.778 | 300.886$^{\alpha}$ | 378.030$^{\alpha}$ |
| | | (7.471) | (7.345) | (45.012) | (56.266) |
| | 30cm, STL | -18.587 | 33.096 | 319.676$^{\alpha}$ | 414.323$^{\alpha}$ |
| | | (7.803) | (8.801) | (67.288) | (90.801) |
| | 45cm, SFL | -16.766 | 37.068 | 351.569$^{\beta}$ | 424.723$^{\beta\dagger}$ |
| | | (6.342) | (6.424) | (45.731) | (57.323) |
| | 45cm, STL | -17.937 | 35.752 | 389.171 | 468.412 |
| | | (4.461) | (5.871) | (71.532) | (70.870) |
| | 60cm, SFL | -18.446 | 38.554 | 397.547$^{\gamma}$ | 472.006$^{\gamma}$ |
| | | (4.356) | (3.942) | (50.068) | (47.567) |
| | 60cm, STL | -17.875 | 36.787 | 405.347$^{\gamma}$ | 483.273 |
| | | (6.918) | (8.187) | (106.754) | (114.625) |
| Knee | 30cm, SFL | 33.723$^{\dagger}$ | 76.069$^{\dagger}$ | 307.292$^{\dagger}$ | 412.077$^{\alpha\dagger}$ |
| | | (7.164) | (11.495) | (45.380) | (32.849) |
| | 30cm, STL | 20.840 | 29.605$^{\alpha}$ | 196.138$^{\alpha}$ | 297.652$^{\alpha}$ |
| | | (4.881) | (11.414) | (51.935) | (85.572) |
| | 45cm, SFL | 35.010$^{\dagger}$ | 74.677$^{\dagger}$ | 327.077$^{\beta\dagger}$ | 456.733$^{\beta\dagger}$ |
| | | (8.055) | (11.164) | (46.321) | (53.210) |
| | 45cm, STL | 22.453$\beta$ | 38.375 | 256.003$^{\beta}$ | 402.464$^{\beta}$ |
| | | (4.412) | (10.501) | (51.452) | (73.879) |
| | 60cm, SFL | 34.452$^{\dagger}$ | 76.387$^{\dagger}$ | 345.373$^{\gamma\dagger}$ | 492.532$^{\gamma}$ |
| | | (7.413) | (12.774) | (43.784) | (48.040) |
| | 60cm, STL | 25.504$^{\gamma}$ | 41.219$^{\gamma}$ | 301.473$^{\gamma}$ | 452.820$^{\gamma}$ |
| | | (8.876) | (9.322) | (49.190) | (72.999) |
| Hip | 30cm, SFL | 28.849$^{\dagger}$ | 85.782$^{\dagger}$ | 206.978$^{\dagger}$ | 364.163$^{\alpha\dagger}$ |
| | | (7.067) | (12.422) | (30.952) | (44.695) |
| | 30cm, STL | 18.783 | 16.115$^{\alpha}$ | 93.998$^{\alpha}$ | 159.451$^{\alpha}$ |
| | | (8.635) | (11.758) | (57.313) | (75.461) |
| | 45cm, SFL | 32.738$^{\dagger}$ | 82.782$^{\dagger}$ | 224.808$^{\beta\dagger}$ | 400.277$^{\beta\dagger}$ |
| | | (7.216) | (12.743) | (34.112) | (51.562) |
| | 45cm, STL | 20.621$^{\beta}$ | 22.091 | 135.182$^{\beta}$ | 222.556 |
| | | (8.156) | (15.998) | (53.766) | (93.583) |
| | 60cm, SFL | 33.314$^{\gamma\dagger}$ | 84.783$^{\dagger}$ | 241.369$^{\gamma\dagger}$ | 423.435$^{\gamma}$ |
| | | (6.0) | (11.107) | (30.649) | (46.106) |
| | 60cm, STL | 23.757$^{\gamma}$ | 23.126$^{\gamma}$ | 168.955$^{\gamma}$ | 205.7454 |
| | | (12.011) | (11.470) | (72.760) | (164.580) |

Note:
$^{\alpha}$. Significant difference between 30 and 45cm for the same landing technique.
$^{\beta}$. Significant difference between 45 and 60cm for the same landing technique.
$^{\gamma}$. Significant difference between 60 and 30cm for the same landing technique.
$\dagger$. Significant difference between SFL and STL at the same height.
Angle and ROM units are in degrees and time unit is in second. Velocity is in deg/s.
The definitions of variables are in Appendix A.

The post-hoc comparison indicated a significant decrease for ContAng, ROM, and ContVel from SFL to STL at each height (Table 2). MaxVel significantly increased from STL to SFL at the two lower heights. A significant increase in ContAng occurred from 45 to 60cm and from 30 to 60cm in STL. ROM showed a significant increase from 30 to 45cm and 30 to 60cm in STL. The post-hoc comparison also indicated a significant increase of ContVel in STL across all heights and a significant increase from 30 to 60cm and 45 to 60cm in SFL. MexVel demonstrated a significant increase across all heights for the same landing technique.

For the hip joint, the ANOVA showed a significant main effect of technique for ContAng, ROM, and ContVel and a significant main effect of height for ContVel and MaxVel. The post-hoc comparison indicated a significant increase for ContAng, ROM, and ContVel from STL to SFL at each height (Table 2). The results showed for MaxVel a significant increase from STL to SFL at 30 and 45cm. The results indicated that ContAng significantly increased in STL from 30 to 60cm and from 45 to 60cm, and in SFL from 30 to 60cm. The hip ROM significantly increased from 30 to 45cm and from 30 to 60cm in STL. The results also revealed that ContVel significantly increased from 30 to 60cm and 45 to 60cm in SFL and across all heights in STL. The results showed that MaxVel significantly increased at each height for SFL and from 30 to 45cm for STL.

## Chapter V

## Discussion

The crux of the study involved using three variables (amplitude, frequency, and damping coefficient) to describe the response of the vastus lateralis muscle under an applied impact. First, let's consider the relationship between the amplitude and frequency by thinking about a wooden bar on a xylophone (28). The "job" of a particular block on the xylophone is to vibrate with a particular frequency and make a sound with a specific pitch. The vibrational frequency of the block is essentially independent of the force with which the block is struck. The use of the block as a musical instrument depends on it reliably making sounds with the same pitch. The natural frequency, or resonance frequency, of the block is determined by its size, density, and damping coefficient (the rate at which the oscillations are damped out). Striking the block harder does not change the pitch (frequency), but it does change the volume (amplitude).

The relationship between frequency, amplitude and force gets more complicated when more than one bar on the xylophone is considered. The blow from the mallet supplies a finite amount of energy for the vibrations. The energy of the vibrations depends on both the frequency and the amplitude ($E = \frac{1}{2}$ mass * frequency$^{\wedge 2}$ * amplitude$^{\wedge 2}$, for harmonic oscillations). This means that two different blocks with different resonance frequencies struck with similar blows, will have different amplitudes.

As for the relationship between damping and frequency, consider a pendulum swaying along at some fixed frequency in the air. Suppose you increase the damping of the pendulum by having it swing in some other medium, like water. The frequency will decrease as the damping tends to slow the motion.

The vastus lateralis muscle is a large, single-joint, pennate muscle, and therefore the imparted vibration is subject to the mechanical properties of the tissue. The mechanical characteristics of the system are governed by muscle geometry, motor unit recruitment, subcutaneous fat, muscle-tendon compliance, intramuscular pressure, and circumference of the muscle complex. Any variation in muscle geometry, motor unit recruitment, or muscle-tendon compliance, which increases tension, will result in increases to the frequency of vibrations. When intramuscular pressure increases, the frequency of vibrations will increase. An increase in muscle circumference will result in a decrease in amplitude. Increases in subcutaneous fat will result in a decrease of frequency. One or more of the above factors may have caused differences in the acceleration signal.

In a pennate muscle, muscle fibers are aligned at a pennate angle to the long axis of the muscle and the angle of pennation can affect the maximal force capability (12). The pattern of motor unit recruitment will alter the muscular force, muscle length, and local fiber orientation. Motor unit recruitment will occur from muscle spindle feedback in less than 100ms if stimulated, which is immediate enough to alter the vibration response. The mean skinfold (SKF) measurement at the attachment site was 9.5 ± 2.6mm. The attachment is inferior to the SKF site of the thigh, however the standard deviation of this measurement is moderately low. Measurement at this site should reduce signal error due to subcutaneous fat. Intramuscular pressure is a result of muscle activation and will increase with an increased number of motor units firing and an increase in activation frequency within those motor units. It has been suggested that high intramuscular pressure is caused by curved muscle fibers exerting an inward pressure

during contraction (11). The muscle circumference will dictate the energy of the vibration.

The mallet strike, delivered by the PI, was considered consistent and reliable through practice, intrinsic feedback (the consistent "feel" of delivering the mallet strike by the investigator), and monitoring the x-axis acceleration curve (26). The only indication of possible inconsistencies in delivery of the mallet strike is the significant increase of amplitude in the x-axis for the squat condition at 35° of knee flexion from all other conditions. However, this increase may be due to the mechanical properties of the soft-tissue complex rather than the investigator's error. Therefore, the results are assumed to be the actual response of the soft-tissue complex in the test conditions.

The mallet strike was delivered perpendicular to the surface of the vastus lateralis and should have the greatest influence on the y-axis acceleration. The mean amplitude in the y-axis acceleration was greater in each condition than the other two axes. The greater amplitude in y-axis is mainly due to the direction of the imparted impact (26). The smallest mean amplitudes were observed in the x-axis and may be due to local fiber orientation. The frequencies between channels showed general stability; however, the frequencies for the z-axis were significantly less than the x-axis for the relaxed conditions. This result may indicate an increased ability of the muscle to reduce the intensity of vibrations along the length of the muscle compared to the width during the non-contraction state. The damping coefficients showed no significant differences among axes or between conditions within their channel. This result is due to a mathematical isotope, whereby the value for $c$ would approach infinity every 20ms.

The difference in amplitude for each respective axis between conditions is most likely due to the different response of the soft-tissue complex to various changes in its structural properties. The x and y axes both showed an increase in amplitude from the relaxed to the contracted conditions indicating an increase in the structural integrity of the complex. In the z-axis acceleration, a decrease of amplitude was noted from the relaxed to the contracted conditions. According to this result and the equation of vibrational energy (E = ½ mass * frequency^2 * amplitude^2), for the energy of the vibration to remain constant, the diameter of the muscle would have to decrease.

The general trend showed an increased frequency from the resting to the squatting conditions. This result is analogous to a guitar string. When tension increases, the number of oscillations in a given time period increases. Wakeling and Nigg (26) reported a significant increase of $f$ and $c$ with increased torque at any joint angle, and an increase of $f$ with an increase in maximum voluntary contraction (MVC) (27). They reported mean $f$ values of 8.2 - 41.3 $s^{-1}$ and mean $c$ values of 8.4 - 61.5 $s^{-1}$ for the quadriceps tissue at the 1st and 4th quartiles of joint torques, and an increase from 9 to 32 Hz between 0% and 100% MVC. For this study, the frequency and coefficients ranged from 21.4 to 300 $s^{-1}$ and from 1.9 to 2226.5 $s^{-1}$, respectively. The data set for frequency showed 10 data points ranging from 150 to 300 $s^{-1}$. Placement of the accelerometer, make of the accelerometer, testing conditions, and the subject population could be factors associated with the increased frequencies. The accelerometer in this study was placed more distally (closer to the vastus lateralis tendon) than done in the Wakeling's and Nigg's studies; this was done to control for motion artifacts during the drop landing conditions. If the amount of tension were greater at the insertion, because muscles only pull towards the

origin, then it would be reasonable to expect the higher *f* values seen in this study. The observation of the data suggested that upper bound value for *c* was approximately 88.8 $s^{-1}$ to 110.3 $s^{-1}$ and subsequent increases in *c* are a result of the isotopes.

The vertical ground reaction force results indicated that F1, F2, LRF1, LRF2, and Impulse significantly increased when knee range of motion was limited for stiff landings (Table1). The VGRF results also showed that F1, F2, LRF1, and Impulse significantly increased as the landing height increased. The results demonstrate an increased loading to the human body as height and stiffness of the lower extremities increases. These results are consistent with prior works in drop landings (4, 5, 31).

The original purpose of this study was to investigate the effects of soft-tissue vibrations of a quadriceps muscle on impact attenuation during landing activity. The data files necessary for such comparisons have been collected and are available for future analysis. However, the most significant problem, which prevented analysis during this study, was signal composition during the drop landings. The two distinct VGRF peaks (F1 & F2) would not allow for calculation of the dominant frequency. Rather, numerous frequencies were discovered and the custom program used for analysis could not determine this component. Another problem posed was the lack of implementation of non-linear least-squares minimization (Levenberg-Marquant method), to determine the best fit of the acceleration data for better estimation of the amplitude, frequency, and damping coefficient.

This study did show increased trends for the frequencies and damping coefficients with increased contraction of the muscle. This study provided pilot work for future study.

Recommendations for future studies of soft-tissue vibrations include:

1. To determine a procedure to analyze acceleration signal that removes the isotope from the damping coefficient.
2. To implement the Levenberg-Marquardt method of non-linear least squares minimization.
3. To determine a procedure to analyze multiple frequencies in a given trial, which can then reduce the signal to only its dominant frequency.
4. To test the possible differences between accelerometers.

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# Appendices

# Appendix A

## Definition of Variables

Definitions of Variables

**Vertical Ground Reaction Force**

| | |
|---|---|
| F1 | First maximum vertical ground reaction force |
| T1 | Time to first maximum vertical ground reaction force |
| F2 | Second maximum vertical ground reaction force |
| T2 | Time to second maximum vertical ground reaction force |
| LRF1 | Loading rate of first maximum vertical ground reaction force\ |
| LRF2 | Loading rate of second maximum vertical ground reaction force |
| Impulse | Impulse of vertical ground reaction force from contact to 100ms |

**Kinematics**

| | |
|---|---|
| ContAng | Joint angle at touchdown |
| Max | Maximum joint angle |
| Tmax | Time to maximum joint angle |
| ROM | Range of motion of joint |
| ContVel | Angular joint velocity at ground contact |
| MaxVel | Angular joint maximum velocity |
| TmaxVel | Time to angular joint maximum velocity |

**Acceleration**

| | |
|---|---|
| FreqX | Frequency of vibration in X-direction |
| CX | Damping Coefficient in X-direction |
| AmpX | Amplitude of vibration in X-direction |
| FreqY | Frequency of vibration in Y -direction |
| CY | Damping Coefficient in Y X-direction |
| AmpY | Amplitude of vibration in Y-direction |
| FreqZ | Frequency of vibration in Z -direction |
| CZ | Damping Coefficient in Z-direction |
| AmpZ | Amplitude of vibration in Z-direction |

INFORMED CONSENT FORM


| | | |
|---|---|---|
| Investigators: | Charles D. Kohstall Jr. | Dr. Song-Ning Zhang |
| Address: | The University of Tennessee<br>Rm. 144, HPER Building<br>1914 Andy Holt Avenue<br>Knoxville, TN 37996 | The University of Tennessee<br>Rm. 337, HPER Building<br>1914 Andy Holt Avenue<br>Knoxville, TN 37996 |
| Phone: | (865) 974-8768 | (865) 974-4716 |


You are invited to participate in a research study entitled, "Relationship between muscle tone and landing biomechanics", which examines the relationship between the muscles on the front of your thigh and impact absorption during a landing activity and back squat.

You are aware that you should be either highly trained or a recreational athlete, have no history of major impairments to your lower extremities, and are injury-free at the time of testing. You must also be able to perform 10 repetitions of a back squat with a weight equal to your body weight. If you decide to participate, you will be asked to attend two testing sessions. The first test session will last approximately 40 minutes and the second test session will last approximately 1 hour. Please wear loose running shorts, and a comfortable short-sleeved shirt or tank top to the test session. At the beginning of the first test session you will fill out an information sheet about your age, height, and sport activity. You will be asked to warm up on a stationary bike for 5 minutes at a self-selected moderate intensity. We will then take anthropometric measurements (approx. 10 minutes) of the lengths and girths of your lower extremities. In the first test session, you will do 10 squats with a weight equal to your body weight to show proficiency in the back squat exercise. You will then be asked to perform 3 back squats with a weight equal to your body weight at knee flexion angles of 30°, 35°, 45°, 65° and 85°, for a total of 15 trials. Each squat at the required position will be held for 10 seconds while a muscle vibration is imparted on your quadriceps muscle with a moderate strike of a rubber reflex-mallet delivered by the primary investigator. The mallet strike will not injure or bruise your muscle, but may feel uncomfortable. Then you will be asked to lie on an athletic training table with you leg positioned at 30° and 35° of knee flexion for 3 trials at each angle and muscle vibrations will be imparted by a similar mallet strike.

You will be asked to return for a second test session with a minimum of two days rest, but no more than 4 days after the first session. You will be asked to warm up on the stationary bike for 5 minutes. At the beginning of each test condition, you will perform 3 landing trials on a force platform. You will then be asked to perform 5 drop landings in each of 6 conditions from an overhead hanging bar onto the force platform for a total of 30 landing trials. The six test conditions include a combinations of landing from 30, 45, and 60cm with soft and stiff landing techniques. Soft landings will utilize deep knee bends upon touch down and stiff landings will utilize less than normal knee flexion. The order of drop heights and landing techniques tested will be randomized. During the test, biomechanics instruments will be used. Some of these instruments will be affixed on your body. None of the instruments will impede your ability to engage in normal and effective

motions during the test. If you have any further questions, interests or concerns about any instrumentation, please feel free to contact the investigator.

Potential risks include an ankle and muscular strain in the lower extremity. Every effort will be made to reduce these risks through proper warm up, sufficient practice before the test, and use of spotters. All tests will be conducted and equipment will be handled by the qualified research personnel in the Biomechanics/Sports Medicine lab. However, these requirements are not beyond those observed in normal weight training or recreational sport activities. The Biomechanics/Sports Medicine Lab has tested more than 120 subjects in several jumping/landing related studies over the past five years. No subject has ever been injured in any fashion during the test sessions. Should any injury occur during the course of testing, standard first aid procedures would be administered as necessary. At least one researcher will be present at each test session who has a basic knowledge of athletic training and/or first aid procedures. In the event a physical injury is suffered as a result of participation in this study, the University of Tennessee does not automatically provide reimbursement for medical care or other compensation. On the other hand, your benefits include the opportunity to learn and receive feedback about individual kinematic, kinetic, and neuromuscular characteristics of landing performance and technique. These characteristics could help improve your skill level or prevent injury. You are welcome to make an appointment to review data from your tests, and/or obtain a copy of the results of this study upon request.

Your participation is entirely voluntary and that your decision whether or not to participate will involve no penalty or loss of benefits to which you are otherwise entitled. Your identity as a subject will be held in strict confidence and any description of your data will be referred to by a subject number only. Any information that is obtained in connection with this study and can be identified as you will remain confidential and will be disclosed only with your permission.

After you have read this informed consent form and all of your questions have been answered, you are requested to sign and date the form below and the attached form that lists individual subject requirements. Your signature indicates that you have read and understand the information provided above, that you willingly agree to participate, that you may withdraw your consent and discontinue participation at any time without penalty or loss of benefits to which you are otherwise entitled.

| Subject Name: | Signature: | Date: |
|---|---|---|
| _____ | _____ | _____ |
| Investigator: | Date: | |
| _____ | _____ | |

**Subject Information**

Project:_____ Test Date:_____

Subject Name:_____ Subject Number:_____

Subject Weight:_____ Subject Height:_____

Age:_____ Sport/Activity:_____

Years Experience:_____ Thigh SKF:_____

| Anthropometric Measurement | | | | |
|---|---|---|---|---|
| Foot | Trial | Prox. Height | Dist. Circum. | Length |
| | 1 | | | |
| | 2 | | | |
| | 3 | | | |
| | Mean | | | |
| | Ankle Mom. | Arm:_____ | | |
| Leg | Trial | Prox. Circum | Dist. Circum. | Length |
| | 1 | | | |
| | 2 | | | |
| | 3 | | | |
| | Mean | | | |
| Thigh | Trial | Prox. Circum | Dist. Circum. | Length |
| | 1 | | | |
| | 2 | | | |
| | 3 | | | |
| | Mean | | | |

## Appendix D

## Vertical Ground Reaction Force Tables

**Table D-1. Means and Standard Deviations of Vertical Ground Reaction Force Data**

| Subject | Condition | F1 | F2 | LRF1 | LRF2 | Imp 100ms |
|---|---|---|---|---|---|---|
| 2 | 30cm, SFL | 9.511 | 20.012 | 537.713 | 217.434 | 1.340 |
| | | (1.646) | (2.651) | (87.547) | (74.827) | (0.216) |
| | 30cm, STL | 23.781 | 47.483 | 1441.395 | 592.275 | 2.779 |
| | | (1.958) | (10.064) | (227.747) | (246.534) | (0.255) |
| | 45cm, SFL | 18.134 | 26.040 | 1226.298 | 290.506 | 1.815 |
| | | (4.003) | (6.196) | (276.853) | (147.244) | (0.292) |
| | 45cm, STL | 31.412 | 45.346 | 2177.850 | 636.716 | 2.849 |
| | | (1.756) | (7.204) | (239.295) | (297.507) | (0.078) |
| | 60cm, SFL | 26.910 | 33.671 | 1946.464 | 499.223 | 2.123 |
| | | (2.468) | (7.023) | (183.099) | (255.333) | (0.161) |
| | 60cm, STL | 37.715 | 45.789 | 2948.249 | 769.497 | 2.914 |
| | | (3.288) | (9.887) | (349.532) | (269.581) | (0.307) |
| 3 | 30cm, SFL | 11.209 | 21.373 | 714.504 | 220.499 | 1.342 |
| | | (2.474) | (3.146) | (344.172) | (63.841) | (0.163) |
| | 30cm, STL | 28.991 | 57.647 | 2225.260 | 890.788 | 3.050 |
| | | (2.432) | (12.269) | (436.431) | (292.607) | (0.129) |
| | 45cm, SFL | 25.312 | 47.322 | 2859.249 | 1362.014 | 2.107 |
| | | (7.957) | (18.714) | (702.374) | (1131.093) | (0.220) |
| | 45cm, STL | 49.665 | 69.833 | 4327.166 | 1275.896 | 3.543 |
| | | (3.931) | (17.168) | (388.768) | (439.426) | (0.416) |
| | 60cm, SFL | 50.076 | 63.037 | 4726.861 | 1519.299 | 2.529 |
| | | (6.357) | (7.825) | (859.783) | (383.404) | (0.231) |
| | 60cm, STL | 71.319 | 92.246 | 6885.002 | 2287.751 | 3.744 |
| | | (12.689) | (14.293) | (1583.421) | (647.852) | (0.192) |
| 4 | 30cm, SFL | 11.644 | 33.144 | 1564.849 | 934.744 | 1.582 |
| | | (1.990) | (4.859) | (755.307) | (595.324) | (0.111) |
| | 30cm, STL | 15.991 | 35.912 | 1162.521 | 582.229 | 1.979 |
| | | (1.420) | (5.80) | (136.161) | (158.873) | (0.106) |
| | 45cm, SFL | 18.616 | 49.549 | 2264.122 | 1685.255 | 2.082 |
| | | (2.550) | (3.726) | (269.491) | (406.029) | (0.106) |
| | 45cm, STL | 24.211 | 47.280 | 2106.610 | 1132.966 | 2.503 |
| | | (1.749) | (3.655) | (46.390) | (169.064) | (0.228) |
| | 60cm, SFL | 24.663 | 67.712 | 3115.023 | 2536.622 | 2.557 |
| | | (2.803) | (7.488) | (295.327) | (847.890) | (0.182) |
| | 60cm, STL | 28.666 | 64.730 | 2871.740 | 1685.656 | 3.075 |
| | | (1.712) | (5.096) | (363.838) | (250.997) | (0.073) |

Note: Force unit is in N/kg and time is in s.
Loading rate unit is in N/kg/s
The definitions of variables are in Appendix A.

**Table D-1 Continued.**

| Subject | Condition | F1 | F2 | LRF1 | LRF2 | Imp 100ms |
|---|---|---|---|---|---|---|
| 5 | 30cm, SFL | 10.219 | 48.194 | 1246.223 | 2128.072 | 1.765 |
| | | (1.823) | (11.387) | (480.380) | (1371.214) | (0.230) |
| | 30cm, STL | 10.776 | 57.597 | 3402.425 | 3500.366 | 2.021 |
| | | (1.482) | (9.716) | (1401.684) | (1081.802) | (0.292) |
| | 45cm, SFL | 15.102 | 67.630 | 2070.180 | 3109.608 | 1.836 |
| | | (1.147) | (7.962) | (199.732) | (875.896) | (0.154) |
| | 45cm, STL | 16.253 | 72.319 | 2057.539 | 3724.155 | 2.270 |
| | | (2.746) | (16.231) | (366.567) | (2089.286) | (0.184) |
| | 60cm, SFL | 28.419 | 63.680 | 2520.680 | 2377.249 | 2.332 |
| | | (2.579) | (15.057) | (485.865) | (1507.017) | (0.210) |
| | 60cm, STL | 28.988 | 83.588 | 2897.026 | 3514.592 | 2.695 |
| | | (3.548) | (13.197) | (140.097) | (1568.642) | (0.149) |
| 6 | 30cm, SFL | 8.292 | 32.299 | 887.067 | 899.905 | 1.466 |
| | | (1.148) | (5.796) | (295.656) | (452.798) | (0.158) |
| | 30cm, STL | 17.467 | 44.463 | 1006.776 | 681.937 | 2.405 |
| | | (2.054) | (10.265) | (87.556) | (284.515) | (0.230) |
| | 45cm, SFL | 18.390 | 46.799 | 1869.224 | 1190.517 | 2.135 |
| | | (1.008) | (8.076) | (352.797) | (445.885) | (0.187) |
| | 45cm, STL | 23.075 | 68.839 | 2644.020 | 1778.471 | 3.392 |
| | | (5.546) | (10.232) | (633.452) | (997.739) | (0.232) |
| | 60cm, SFL | 27.798 | 59.209 | 2886.795 | 1679.505 | 2.596 |
| | | (2.156) | (6.337) | (277.159) | (304.155) | (0.126) |
| | 60cm, STL | 21.882 | 87.619 | 3407.392 | 3379.543 | 3.410 |
| | | (4.418) | (13.530) | (574.961) | (1322.876) | (0.214) |
| 7 | 30cm, SFL | 8.944 | 23.247 | 579.664 | 234.456 | 1.166 |
| | | (0.692) | (0.918) | (97.655) | (27.139) | (0.057) |
| | 30cm, STL | 20.994 | 58.915 | 1351.468 | 948.727 | 2.969 |
| | | (1.571) | (3.492) | (185.471) | (152.857) | (0.132) |
| | 45cm, SFL | 15.979 | 30.405 | 998.918 | 354.341 | 1.714 |
| | | (3.025) | (6.338) | (160.163) | (111.393) | (0.257) |
| | 45cm, STL | 31.243 | 79.387 | 2449.545 | 1808.587 | 3.574 |
| | | (1.092) | (6.864) | (198.142) | (374.829) | (0.102) |
| | 60cm, SFL | 30.886 | 41.548 | 2592.097 | 783.920 | 2.371 |
| | | (3.880) | (10.543) | (591.094) | (422.256) | (0.253) |
| | 60cm, STL | 37.069 | 85.021 | 3232.757 | 2305.861 | 3.759 |
| | | (2.094) | (8.673) | (142.483) | (362.340) | (0.259) |

Note: Force unit is in N/kg and time is in s.
Loading rate unit is in N/kg/s
The definitions of variables are in Appendix A.

**Table D-1 Continued.**

| Subject | Condition | F1 | F2 | LRF1 | LRF2 | Imp 100ms |
|---|---|---|---|---|---|---|
| 8 | 30cm, SFL | 5.052 | 19.301 | 336.191 | 229.551 | 1.083 |
| | | (1.373) | (2.224) | (78.896) | (26.575) | (0.149) |
| | 30cm, STL | 14.406 | 56.731 | 806.403 | 853.662 | 2.909 |
| | | (1.50) | (4.366) | (74.253) | (210.306) | (0.094) |
| | 45cm, SFL | 14.983 | 29.079 | 1176.002 | 303.672 | 2.011 |
| | | (2.284) | (2.778) | (239.770) | (100.443) | (0.171) |
| | 45cm, STL | 17.160 | 72.439 | 1529.841 | 1752.815 | 3.135 |
| | | (2.149) | (8.209) | (424.733) | (320.591) | (0.241) |
| | 60cm, SFL | 19.301 | 33.751 | 1476.785 | 496.033 | 2.122 |
| | | (2.701) | (9.004) | (267.735) | (318.674) | (0.155) |
| | 60cm, STL | 31.856 | 73.251 | 2363.615 | 1601.525 | 3.633 |
| | | (3.549) | (7.558) | (223.857) | (442.337) | (0.080) |
| 9 | 30cm, SFL | 6.858 | 28.010 | 522.776 | 522.152 | 1.287 |
| | | (0.611) | (4.927) | (62.064) | (198.147) | (0.074) |
| | 30cm, STL | 9.996 | 52.888 | 838.460 | 1060.750 | 2.364 |
| | | (0.570) | (5.670) | (196.484) | (174.657) | (0.117) |
| | 45cm, SFL | 13.374 | 41.639 | 887.719 | 726.845 | 2.208 |
| | | (1.159) | (9.168) | (94.429) | (261.609) | (0.146) |
| | 45cm, STL | 17.672 | 57.866 | 1227.563 | 1140.651 | 2.812 |
| | | (1.431) | (10.656) | (266.957) | (423.260) | (0.193) |
| | 60cm, SFL | 20.297 | 50.167 | 1419.648 | 1004.772 | 2.462 |
| | | (1.656) | (3.765) | (227.736) | (169.642) | (0.160) |
| | 60cm, STL | 26.581 | 63.946 | 1770.214 | 1329.396 | 3.166 |
| | | (1.496) | (4.339) | (238.582) | (97.758) | (0.163) |
| 10 | 30cm, SFL | 10.579 | 24.297 | 693.556 | 262.948 | 1.599 |
| | | (1.164) | (3.434) | (140.740) | (51.659) | (0.229) |
| | 30cm, STL | 19.509 | 48.138 | 1223.488 | 616.564 | 3.385 |
| | | (1.738) | (2.803) | (288.819) | (97.186) | (0.124) |
| | 45cm, SFL | 18.531 | 24.786 | 1430.128 | 338.649 | 1.811 |
| | | (3.401) | (2.588) | (290.638) | (128.438) | (0.167) |
| | 45cm, STL | 32.810 | 60.576 | 2433.643 | 784.044 | 3.801 |
| | | (1.747) | (5.952) | (177.848) | (168.914) | (0.157) |
| | 60cm, SFL | 25.620 | 34.003 | 2006.425 | 528.086 | 2.278 |
| | | (4.381) | (10.664) | (476.167) | (241.786) | (0.417) |
| | 60cm, STL | 37.565 | 61.363 | 2201.599 | 761.124 | 4.017 |
| | | (2.406) | (6.861) | (150.744) | (292.020) | (0.188) |

Note: Force unit is in N/kg and time is in s.
Loading rate unit is in N/kg/s
The definitions of variables are in Appendix A.

**Table D-1 Continued.**

| Subject | Condition | F1 | F2 | LRF1 | LRF2 | Imp 100ms |
|---|---|---|---|---|---|---|
| 11 | 30cm, SFL | 10.026 | 6.540 | 732.806 | 603.369 | 1.597 |
| | | (1.293) | (0.893) | (82.770) | (105.484) | (0.084) |
| | 30cm, STL | 15.832 | 12.218 | 1115.807 | 1151.215 | 2.891 |
| | | (1.328) | (1.581) | (112.190) | (128.235) | (0.197) |
| | 45cm, SFL | 17.114 | 9.603 | 1426.133 | 983.400 | 2.096 |
| | | (1.552) | (1.353) | (191.337) | (462.694) | (0.145) |
| | 45cm, STL | 28.188 | 20.523 | 1700.392 | 1452.046 | 3.713 |
| | | (2.198) | (1.626) | (114.843) | (307.928) | (0.060) |
| | 60cm, SFL | 25.483 | 12.192 | 1845.203 | 1449.109 | 2.551 |
| | | (2.203) | (1.292) | (154.964) | (635.740) | (0.114) |
| | 60cm, STL | 36.711 | 20.767 | 2301.134 | 1747.025 | 3.884 |
| | | (1.699) | (1.166) | (164.226) | (294.422) | (0.140) |

Note: Force unit is in N/kg and time is in s.
Loading rate unit is in N/kg/s
The definitions of variables are in Appendix A.

# Appendix E

## Kinematic Tables

**Table E-1. Means and Standard Deviations of Subject 2 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -19.498 | 45.301 | 318.892 | 395.588 |
| | | (2.343) | (3.007) | (23.932) | (8.374) |
| | Knee | 35.430 | 73.947 | 340.914 | 382.263 |
| | | (2.950) | (4.435) | (22.781) | (20.414) |
| | Hip | 24.251 | 82.625 | 195.121 | 288.104 |
| | | (5.558) | (4.721) | (17.518) | (19.206) |
| 30cm, STL | Ankle | -21.073 | 33.608 | 389.779 | 477.635 |
| | | (2.511) | (6.150) | (42.955) | (83.483) |
| | Knee | 21.024 | 19.286 | 228.490 | 245.058 |
| | | (1.351) | (2.704) | (19.917) | (18.346) |
| | Hip | 14.561 | 8.063 | 80.597 | 96.325 |
| | | (1.431) | (1.426) | (15.385) | (13.164) |
| 45cm, SFL | Ankle | -17.538 | 39.150 | 384.336 | 443.519 |
| | | (3.821) | (5.220) | (22.415) | (51.062) |
| | Knee | 38.769 | 83.533 | 403.622 | 454.821 |
| | | (4.572) | (10.669) | (55.490) | (23.190) |
| | Hip | 26.287 | 93.474 | 245.895 | 345.104 |
| | | (4.735) | (4.704) | (41.389) | (10.137) |
| 45cm, STL | Ankle | -17.438 | 39.086 | 469.940 | 526.349 |
| | | (0.782) | (2.292) | (33.274) | (24.240) |
| | Knee | 28.571 | 32.616 | 343.145 | 359.659 |
| | | (1.980) | (3.586) | (21.293) | (17.318) |
| | Hip | 17.563 | 15.076 | 165.469 | 176.569 |
| | | (2.817) | (3.584) | (16.702) | (14.443) |
| 60cm, SFL | Ankle | -16.115 | 40.005 | 438.207 | 506.201 |
| | | (0.739) | (2.569) | (33.138) | (36.579) |
| | Knee | 39.998 | 93.845 | 410.215 | 511.957 |
| | | (4.965) | (3.863) | (22.338) | (7.365) |
| | Hip | 28.948 | 95.184 | 263.631 | 396.671 |
| | | (6.009) | (2.332) | (12.743) | (12.854) |
| 60cm, STL | Ankle | -17.979 | 42.595 | 449.489 | 523.960 |
| | | (1.745) | (2.678) | (29.199) | (22.261) |
| | Knee | 30.920 | 45.850 | 390.997 | 454.501 |
| | | (2.206) | (7.930) | (26.688) | (23.490) |
| | Hip | 22.375 | 26.008 | 220.082 | 267.601 |
| | | (2.778) | (5.917) | (20.151) | (27.880) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-2. Means and Standard Deviations of Subject 3 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -20.756 | 37.054 | 287.991 | 366.603 |
| | | (3.381) | (4.382) | (30.813) | (30.764) |
| | Knee | 36.061 | 77.900 | 292.975 | 382.836 |
| | | (3.697) | (7.415) | (17.884) | (17.313) |
| | Hip | 25.437 | 106.027 | 207.987 | 415.579 |
| | | (2.292) | (3.540) | (18.971) | (35.867) |
| 30cm, STL | Ankle | -22.702 | 29.652 | 320.011 | 402.907 |
| | | (1.550) | (4.551) | (29.989) | (13.606) |
| | Knee | 16.283 | 13.588 | 121.422 | 160.922 |
| | | (3.748) | (7.457) | (46.602) | (74.662) |
| | Hip | 14.160 | 9.091 | 65.562 | 107.496 |
| | | (3.408) | (4.110) | (23.366) | (44.426) |
| 45cm, SFL | Ankle | -14.886 | 29.472 | 297.603 | 347.085 |
| | | (5.584) | (6.683) | (34.912) | (34.246) |
| | Knee | 38.148 | 66.409 | 295.166 | 403.283 |
| | | (3.604) | (5.80) | (45.616) | (52.142) |
| | Hip | 34.872 | 92.898 | 255.212 | 429.979 |
| | | (5.338) | (2.964) | (44.322) | (24.177) |
| 45cm, STL | Ankle | -19.472 | 31.276 | 411.479 | 468.780 |
| | | (1.603) | (2.904) | (56.505) | (39.533) |
| | Knee | 16.257 | 28.789 | 200.919 | 338.708 |
| | | (3.272) | (2.687) | (15.686) | (21.194) |
| | Hip | 13.676 | 16.690 | 114.992 | 199.179 |
| | | (1.241) | (3.136) | (20.193) | (29.097) |
| 60cm, SFL | Ankle | -17.241 | 33.879 | 388.590 | 434.781 |
| | | (1.654) | (4.098) | (38.823) | (46.449) |
| | Knee | 35.307 | 66.825 | 304.389 | 448.811 |
| | | (3.875) | (8.914) | (34.297) | (35.967) |
| | Hip | 30.750 | 94.296 | 260.716 | 469.220 |
| | | (6.237) | (12.059) | (25.160) | (26.499) |
| 60cm, STL | Ankle | -21.038 | 37.169 | 400.615 | 480.657 |
| | | (1.373) | (1.242) | (41.946) | (34.095) |
| | Knee | 20.049 | 37.826 | 256.373 | 432.927 |
| | | (2.117) | (4.149) | (22.711) | (35.474) |
| | Hip | 15.739 | 17.659 | 132.517 | 220.748 |
| | | (1.518) | (5.256) | (25.007) | (49.318) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-3. Means and Standard Deviations of Subject 4 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -13.391 | 36.861 | 332.573 | 392.085 |
| | | (8.423) | (8.852) | (79.218) | (102.433) |
| | Knee | 23.926 | 82.135 | 289.925 | 430.083 |
| | | (4.249) | (7.419) | (40.841) | (36.110) |
| | Hip | 26.975 | 87.324 | 210.732 | 397.410 |
| | | (7.586) | (8.740) | (46.166) | (16.767) |
| 30cm, STL | Ankle | -20.070 | 41.888 | 387.065 | 483.760 |
| | | (1.956) | (1.728) | (37.249) | (53.710) |
| | Knee | 19.903 | 42.494 | 218.206 | 338.427 |
| | | (0.630) | (1.777) | (19.960) | (10.477) |
| | Hip | 19.261 | 29.389 | 103.300 | 240.355 |
| | | (2.203) | (2.713) | (23.263) | (10.818) |
| 45cm, SFL | Ankle | -19.215 | 42.199 | 390.833 | 453.073 |
| | | (3.518) | (2.945) | (23.335) | (28.062) |
| | Knee | 22.022 | 77.155 | 319.844 | 508.655 |
| | | (2.529) | (3.330) | (47.322) | (42.931) |
| | Hip | 31.488 | 76.841 | 239.930 | 412.828 |
| | | (3.626) | (3.916) | (35.370) | (27.387) |
| 45cm, STL | Ankle | -21.215 | 42.792 | 419.760 | 504.912 |
| | | (2.870) | (2.595) | (63.598) | (47.070) |
| | Knee | 20.387 | 51.710 | 280.898 | 466.535 |
| | | (1.511) | (7.011) | (26.649) | (46.165) |
| | Hip | 25.839 | 36.282 | 176.399 | 343.134 |
| | | (1.587) | (4.397) | (25.713) | (33.059) |
| 60cm, SFL | Ankle | -18.612 | 37.575 | 422.948 | 468.298 |
| | | (2.894) | (3.953) | (70.454) | (63.423) |
| | Knee | 25.691 | 74.938 | 364.512 | 488.024 |
| | | (1.122) | (6.003) | (31.460) | (13.825) |
| | Hip | 36.847 | 80.481 | 271.219 | 421.531 |
| | | (2.238) | (4.282) | (26.862) | (29.681) |
| 60cm, STL | Ankle | -18.973 | 39.241 | 447.175 | 504.412 |
| | | (2.215) | (2.880) | (48.874) | (54.081) |
| | Knee | 22.301 | 47.947 | 326.041 | 484.584 |
| | | (1.702) | (3.973) | (24.221) | (31.492) |
| | Hip | 26.963 | 38.162 | 220.624 | 374.230 |
| | | (1.847) | (4.228) | (22.192) | (15.813) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-4. Means and Standard Deviations of Subject 5 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -8.6 | 23.8 | 244.4 | 285.3 |
| | | (5.179) | (5.243) | (50.831) | (56.837) |
| | Knee | 28.81 | 67.21 | 277.88 | 407.63 |
| | | (3.630) | (8.163) | (29.099) | (21.166) |
| | Hip | 32.5 | 87.8 | 234.5 | 355.8 |
| | | (7.677) | (9.144) | (20.204) | (22.830) |
| 30cm, STL | Ankle | -0.8 | 18.5 | 220.0 | 235.5 |
| | | (1.267) | (2.116) | (38.774) | (32.032) |
| | Knee | 30.10 | 45.23 | 302.14 | 418.08 |
| | | (2.218) | (6.967) | (34.545) | (25.378) |
| | Hip | 35.7 | 41.5 | 246.2 | 307.3 |
| | | (2.142) | (10.731) | (19.537) | (18.254) |
| 45cm, SFL | Ankle | -11.4 | 27.6 | 315.4 | 342.7 |
| | | (1.232) | (1.582) | (49.017) | (44.523) |
| | Knee | 28.61 | 65.97 | 317.22 | 473.72 |
| | | (1.541) | (2.469) | (15.107) | (27.051) |
| | Hip | 39.2 | 78.1 | 244.0 | 391.4 |
| | | (5.455) | (3.159) | (11.931) | (36.046) |
| 45cm, STL | Ankle | -11.3 | 26.9 | 296.5 | 329.6 |
| | | (2.966) | (3.134) | (40.923) | (37.917) |
| | Knee | 26.49 | 48.26 | 302.12 | 461.95 |
| | | (2.474) | (3.435) | (30.609) | (35.305) |
| | Hip | 34.1 | 45.8 | 237.8 | 343.5 |
| | | (7.231) | (8.815) | (18.724) | (38.262) |
| 60cm, SFL | Ankle | -22.2 | 37.2 | 419.3 | 482.9 |
| | | (1.789) | (1.736) | (60.867) | (31.307) |
| | Knee | 24.57 | 64.62 | 326.06 | 485.27 |
| | | (1.631) | (1.353) | (10.160) | (22.116) |
| | Hip | 34.4 | 79.6 | 225.8 | 393.0 |
| | | (5.802) | (6.281) | (32.965) | (26.692) |
| 60cm, STL | Ankle | -14.7 | 29.9 | 323.0 | 383.2 |
| | | (17.145) | (16.963) | (175.907) | (209.217) |
| | Knee | 33.55 | 37.95 | 295.36 | 433.30 |
| | | (25.705) | (20.043) | (37.672) | (115.911) |
| | Hip | 44.0 | 37.6 | 221.2 | 330.0 |
| | | (22.912) | (16.821) | (91.988) | (40.882) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-5. Means and Standard Deviations of Subject 6 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -21.234 | 34.956 | 287.404 | 363.610 |
| | | (8.372) | (6.101) | (35.172) | (51.380) |
| | Knee | 28.151 | 76.610 | 269.671 | 441.692 |
| | | (3.80) | (8.852) | (38.087) | (46.727) |
| | Hip | 41.891 | 89.186 | 202.998 | 398.878 |
| | | (3.351) | (4.014) | (31.653) | (17.969) |
| 30cm, STL | Ankle | -29.113 | 43.992 | 352.051 | 476.114 |
| | | (3.118) | (4.745) | (47.708) | (68.669) |
| | Knee | 21.634 | 36.656 | 186.546 | 318.570 |
| | | (1.888) | (10.777) | (40.454) | (70.529) |
| | Hip | 32.265 | 20.469 | 81.790 | 200.960 |
| | | (3.768) | (6.50) | (30.948) | (52.406) |
| 45cm, SFL | Ankle | -24.074 | 40.154 | 360.156 | 461.591 |
| | | (3.928) | (3.697) | (30.104) | (30.953) |
| | Knee | 27.905 | 73.198 | 311.006 | 511.866 |
| | | (0.932) | (2.361) | (22.183) | (14.480) |
| | Hip | 39.891 | 92.878 | 210.449 | 457.401 |
| | | (3.787) | (6.128) | (29.342) | (20.184) |
| 45cm, STL | Ankle | -18.963 | 35.032 | 327.951 | 422.067 |
| | | (6.802) | (6.376) | (99.704) | (94.139) |
| | Knee | 24.469 | 50.584 | 243.239 | 503.814 |
| | | (3.807) | (12.776) | (47.196) | (46.067) |
| | Hip | 32.657 | 36.865 | 150.509 | 324.439 |
| | | (4.041) | (32.353) | (45.080) | (108.227) |
| 60cm, SFL | Ankle | -27.678 | 41.485 | 378.253 | 489.149 |
| | | (2.874) | (2.446) | (41.937) | (34.494) |
| | Knee | 27.441 | 81.173 | 330.987 | 567.375 |
| | | (2.567) | (5.561) | (21.990) | (26.236) |
| | Hip | 42.061 | 93.720 | 235.467 | 487.156 |
| | | (3.792) | (5.348) | (22.802) | (32.840) |
| 60cm, STL | Ankle | -7.225 | 23.163 | 195.982 | 268.438 |
| | | (8.383) | (9.266) | (77.688) | (108.797) |
| | Knee | 28.262 | 46.301 | 366.627 | 535.433 |
| | | (2.979) | (9.233) | (32.321) | (48.277) |
| | Hip | 39.517 | 33.808 | 292.968 | 1335.658 |
| | | (3.0) | (5.802) | (52.203) | (2137.105) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-6. Means and Standard Deviations of Subject 7 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -25.790 | 47.385 | 296.412 | 437.984 |
| | | (1.836) | (2.921) | (36.357) | (44.675) |
| | Knee | 31.652 | 74.883 | 353.143 | 421.367 |
| | | (3.480) | (4.191) | (27.926) | (12.902) |
| | Hip | 24.400 | 99.569 | 207.695 | 375.862 |
| | | (2.307) | (3.677) | (13.601) | (16.261) |
| 30cm, STL | Ankle | -23.214 | 42.561 | 343.351 | 475.476 |
| | | (1.694) | (3.074) | (59.248) | (51.803) |
| | Knee | 16.065 | 35.706 | 185.297 | 363.842 |
| | | (1.498) | (2.281) | (21.140) | (31.048) |
| | Hip | 15.446 | 9.380 | 66.701 | 114.522 |
| | | (1.732) | (1.186) | (5.787) | (13.327) |
| 45cm, SFL | Ankle | -25.084 | 43.852 | 345.899 | 478.453 |
| | | (2.542) | (5.054) | (33.212) | (36.233) |
| | Knee | 34.902 | 81.982 | 367.939 | 446.668 |
| | | (3.167) | (8.228) | (27.059) | (44.203) |
| | Hip | 25.240 | 97.930 | 207.830 | 444.517 |
| | | (3.407) | (3.080) | (18.569) | (19.724) |
| 45cm, STL | Ankle | -21.951 | 42.182 | 419.061 | 521.794 |
| | | (1.837) | (2.50) | (34.078) | (41.018) |
| | Knee | 18.443 | 40.125 | 223.420 | 432.896 |
| | | (2.163) | (1.628) | (14.963) | (22.950) |
| | Hip | 17.517 | 10.485 | 82.804 | 138.374 |
| | | (1.153) | (1.217) | (10.278) | (14.150) |
| 60cm, SFL | Ankle | -19.483 | 42.412 | 422.634 | 514.651 |
| | | (1.916) | (0.910) | (37.211) | (28.819) |
| | Knee | 34.556 | 73.677 | 360.038 | 512.804 |
| | | (1.917) | (4.418) | (61.649) | (24.326) |
| | Hip | 28.942 | 97.517 | 226.771 | 457.503 |
| | | (2.830) | (4.641) | (27.147) | (22.585) |
| 60cm, STL | Ankle | -19.816 | 42.494 | 448.799 | 539.192 |
| | | (2.474) | (3.540) | (43.250) | (22.531) |
| | Knee | 21.206 | 46.673 | 291.869 | 528.703 |
| | | (2.362) | (4.540) | (23.321) | (52.759) |
| | Hip | 20.830 | 14.169 | 127.447 | 199.081 |
| | | (1.780) | (3.328) | (23.817) | (36.922) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-7. Means and Standard Deviations of Subject 8 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -20.737 | 34.432 | 341.685 | 398.685 |
| | | (2.921) | (2.489) | (31.454) | (32.280) |
| | Knee | 35.668 | 85.219 | 389.413 | 427.908 |
| | | (4.053) | (11.004) | (33.830) | (46.678) |
| | Hip | 25.756 | 84.823 | 245.736 | 364.317 |
| | | (4.558) | (7.799) | (17.599) | (28.657) |
| 30cm, STL | Ankle | -20.356 | 34.530 | 348.642 | 464.810 |
| | | (7.003) | (5.305) | (26.705) | (54.866) |
| | Knee | 16.839 | 24.611 | 179.084 | 255.705 |
| | | (3.190) | (2.831) | (20.571) | (38.038) |
| | Hip | 8.283 | 7.422 | 56.716 | 90.931 |
| | | (1.318) | (3.387) | (20.206) | (38.394) |
| 45cm, SFL | Ankle | -16.548 | 33.486 | 336.729 | 410.315 |
| | | (4.222) | (4.398) | (13.362) | (30.936) |
| | Knee | 42.171 | 74.767 | 303.329 | 371.709 |
| | | (5.747) | (8.094) | (24.338) | (52.097) |
| | Hip | 41.268 | 75.820 | 225.801 | 343.635 |
| | | (9.205) | (3.298) | (13.449) | (31.234) |
| 45cm, STL | Ankle | -13.282 | 32.152 | 447.293 | 489.585 |
| | | (2.828) | (2.219) | (33.674) | (35.860) |
| | Knee | 19.014 | 33.036 | 282.733 | 393.621 |
| | | (2.141) | (3.892) | (22.810) | (42.665) |
| | Hip | 13.839 | 13.967 | 128.438 | 178.325 |
| | | (1.836) | (3.976) | (31.041) | (43.661) |
| 60cm, SFL | Ankle | -14.377 | 33.508 | 411.325 | 470.670 |
| | | (2.876) | (1.670) | (50.539) | (42.855) |
| | Knee | 44.079 | 77.430 | 333.057 | 426.850 |
| | | (3.442) | (7.996) | (29.706) | (15.237) |
| | Hip | 39.741 | 75.265 | 255.417 | 374.837 |
| | | (3.622) | (3.282) | (11.390) | (32.521) |
| 60cm, STL | Ankle | -19.239 | 38.161 | 431.027 | 514.020 |
| | | (1.765) | (3.189) | (35.956) | (35.518) |
| | Knee | 20.223 | 40.321 | 274.221 | 450.167 |
| | | (2.009) | (1.008) | (19.616) | (23.429) |
| | Hip | 15.042 | 18.326 | 149.325 | 226.462 |
| | | (1.680) | (3.646) | (15.358) | (47.396) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-8. Means and Standard Deviations of Subject 9 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -14.753 | 36.165 | 281.648 | 379.482 |
| | | (1.412) | (2.082) | (16.669) | (15.618) |
| | Knee | 29.661 | 58.219 | 293.947 | 406.100 |
| | | (1.872) | (4.545) | (16.824) | (31.066) |
| | Hip | 30.564 | 63.096 | 201.543 | 331.079 |
| | | (6.308) | (5.135) | (24.819) | (22.481) |
| 30cm, STL | Ankle | -20.330 | 29.390 | 243.340 | 355.115 |
| | | (1.738) | (3.866) | (58.225) | (53.005) |
| | Knee | 17.136 | 28.447 | 199.534 | 316.724 |
| | | (1.715) | (2.089) | (17.288) | (30.191) |
| | Hip | 18.882 | 14.307 | 97.407 | 172.114 |
| | | (2.501) | (1.537) | (16.229) | (20.305) |
| 45cm, SFL | Ankle | -21.034 | 42.814 | 354.086 | 454.729 |
| | | (2.021) | (2.20) | (35.543) | (42.227) |
| | Knee | 22.995 | 44.254 | 234.217 | 354.889 |
| | | (12.097) | (23.644) | (122.526) | (181.885) |
| | Hip | 29.262 | 51.571 | 171.666 | 339.209 |
| | | (4.891) | (0.582) | (28.731) | (21.451) |
| 45cm, STL | Ankle | -22.563 | 40.060 | 354.881 | 481.737 |
| | | (2.436) | (2.856) | (61.260) | (50.716) |
| | Knee | 20.193 | 41.622 | 257.656 | 418.761 |
| | | (0.775) | (6.229) | (43.214) | (45.011) |
| | Hip | 22.883 | 23.719 | 128.119 | 247.522 |
| | | (1.955) | (5.449) | (27.837) | (44.030) |
| 60cm, SFL | Ankle | -18.646 | 38.562 | 395.227 | 461.124 |
| | | (1.755) | (2.388) | (9.220) | (15.966) |
| | Knee | 30.882 | 58.584 | 336.918 | 444.971 |
| | | (2.399) | (4.418) | (22.571) | (28.079) |
| | Hip | 31.083 | 66.977 | 216.791 | 388.807 |
| | | (3.639) | (3.554) | (28.312) | (24.230) |
| 60cm, STL | Ankle | -21.820 | 38.250 | 440.078 | 510.973 |
| | | (0.960) | (2.271) | (40.986) | (43.912) |
| | Knee | 22.116 | 40.732 | 281.342 | 431.967 |
| | | (1.049) | (2.035) | (20.161) | (32.048) |
| | Hip | 22.296 | 20.541 | 134.275 | 235.757 |
| | | (1.690) | (2.912) | (24.783) | (29.501) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-9. Means and Standard Deviations of Subject 10 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -11.976 | 36.821 | 291.000 | 368.862 |
| | | (3.272) | (2.472) | (16.774) | (14.650) |
| | Knee | 42.223 | 94.821 | 282.181 | 417.768 |
| | | (2.638) | (5.244) | (18.317) | (15.491) |
| | Hip | 25.211 | 76.716 | 157.806 | 315.097 |
| | | (2.773) | (2.136) | (13.947) | (10.891) |
| 30cm, STL | Ankle | -15.285 | 24.151 | 279.747 | 358.500 |
| | | (2.428) | (5.602) | (50.924) | (68.331) |
| | Knee | 23.667 | 17.452 | 161.779 | 202.568 |
| | | (0.911) | (4.478) | (22.117) | (46.511) |
| | Hip | 14.032 | 7.266 | 54.908 | 93.126 |
| | | (2.862) | (2.723) | (23.936) | (32.313) |
| 45cm, SFL | Ankle | -11.963 | 38.072 | 378.018 | 428.906 |
| | | (3.242) | (2.711) | (67.643) | (62.014) |
| | Knee | 44.799 | 92.314 | 352.218 | 488.097 |
| | | (1.724) | (5.217) | (21.767) | (21.250) |
| | Hip | 29.015 | 78.324 | 211.474 | 363.148 |
| | | (4.167) | (2.928) | (7.136) | (13.704) |
| 45cm, STL | Ankle | -15.962 | 30.982 | 374.765 | 457.791 |
| | | (1.906) | (2.530) | (51.919) | (57.460) |
| | Knee | 25.030 | 23.819 | 199.813 | 268.432 |
| | | (1.392) | (2.488) | (24.490) | (29.333) |
| | Hip | 11.543 | 8.603 | 60.243 | 108.269 |
| | | (3.054) | (3.722) | (20.920) | (44.646) |
| 60cm, SFL | Ankle | -14.524 | 38.433 | 339.175 | 405.368 |
| | | (2.50) | (3.358) | (38.766) | (44.345) |
| | Knee | 44.186 | 97.451 | 369.303 | 529.014 |
| | | (3.172) | (7.062) | (60.716) | (45.026) |
| | Hip | 30.996 | 79.716 | 237.020 | 384.327 |
| | | (2.852) | (6.324) | (43.746) | (29.408) |
| 60cm, STL | Ankle | -18.915 | 37.434 | 511.004 | 588.985 |
| | | (1.852) | (2.121) | (57.424) | (37.081) |
| | Knee | 28.973 | 27.642 | 264.280 | 320.977 |
| | | (1.236) | (6.038) | (18.564) | (36.034) |
| | Hip | 13.570 | 10.757 | 77.535 | 130.592 |
| | | (1.312) | (3.424) | (27.487) | (35.284) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

**Table E-10. Means and Standard Deviations of Subject 11 Kinematic Data.**

| Condition | Joint | ContAng | ROM | ContVel | MaxVel |
|---|---|---|---|---|---|
| 30cm, SFL | Ankle | -19.498 | 45.301 | 318.892 | 395.588 |
| | | (2.343) | (3.007) | (23.932) | (8.374) |
| | Knee | 35.430 | 73.947 | 340.914 | 382.263 |
| | | (2.950) | (4.435) | (22.781) | (20.414) |
| | Hip | 24.251 | 82.625 | 195.121 | 288.104 |
| | | (5.558) | (4.721) | (17.518) | (19.206) |
| 30cm, STL | Ankle | -21.073 | 33.608 | 389.779 | 477.635 |
| | | (2.511) | (6.150) | (42.955) | (83.483) |
| | Knee | 21.024 | 19.286 | 228.490 | 245.058 |
| | | (1.351) | (2.704) | (19.917) | (18.346) |
| | Hip | 14.561 | 8.063 | 80.597 | 96.325 |
| | | (1.431) | (1.426) | (15.385) | (13.164) |
| 45cm, SFL | Ankle | -17.538 | 39.150 | 384.336 | 443.519 |
| | | (3.821) | (5.220) | (22.415) | (51.062) |
| | Knee | 38.769 | 83.533 | 403.622 | 454.821 |
| | | (4.572) | (10.669) | (55.490) | (23.190) |
| | Hip | 26.287 | 93.474 | 245.895 | 345.104 |
| | | (4.735) | (4.704) | (41.389) | (10.137) |
| 45cm, STL | Ankle | -17.438 | 39.086 | 469.940 | 526.349 |
| | | (0.782) | (2.292) | (33.274) | (24.240) |
| | Knee | 28.571 | 32.616 | 343.145 | 359.659 |
| | | (1.980) | (3.586) | (21.293) | (17.318) |
| | Hip | 17.563 | 15.076 | 165.469 | 176.569 |
| | | (2.817) | (3.584) | (16.702) | (14.443) |
| 60cm, SFL | Ankle | -16.115 | 40.005 | 438.207 | 506.201 |
| | | (0.739) | (2.569) | (33.138) | (36.579) |
| | Knee | 39.998 | 93.845 | 410.215 | 511.957 |
| | | (4.965) | (3.863) | (22.338) | (7.365) |
| | Hip | 28.948 | 95.184 | 263.631 | 396.671 |
| | | (6.009) | (2.332) | (12.743) | (12.854) |
| 60cm, STL | Ankle | -17.979 | 42.595 | 449.489 | 523.960 |
| | | (1.745) | (2.678) | (29.199) | (22.261) |
| | Knee | 30.920 | 45.850 | 390.997 | 454.501 |
| | | (2.206) | (7.930) | (26.688) | (23.490) |
| | Hip | 22.375 | 26.008 | 220.082 | 267.601 |
| | | (2.778) | (5.917) | (20.151) | (27.880) |

Note: Angle and ROM units are in degrees and time unit is in seconds.
Velocity unit is deg/s.
The definitions of variables are in Appendix A.

## Appendix F

## Acceleration Tables

**Table F-1. Means and Standard Deviations of Acceleration Data for Individual Subjects**

| Subject | Condition | FreqX | CX | AmpX | FreqY | Cy | AmpY | FreqZ | CZ | AmpZ |
|---|---|---|---|---|---|---|---|---|---|---|
| 2 | 30° Relax | 79.299 | 15.705 | 2.725 | 51.449 | 70.582 | 3.910 | 24.837 | 28.006 | 4.529 |
| | | (11.480) | (6.050) | (0.451) | (1.255) | (82.917) | (0.593) | (0.602) | (6.647) | (0.340) |
| | 35° Relax | 79.365 | 17.666 | 4.052 | 58.182 | 22.580 | 5.966 | 25.717 | 31.222 | 5.557 |
| | | (10.997) | (7.647) | (1.166) | (3.149) | (4.682) | (0.912) | (0.320) | (16.041) | (0.899) |
| | 30° Squat | 66.019 | 25.931 | 2.586 | 59.671 | 44.216 | 14.243 | 44.444 | 30.233 | 3.373 |
| | | (19.974) | (2.071) | (0.451) | (4.523) | (7.960) | (1.415) | (0.0) | (9.824) | (0.328) |
| | 35° Squat | 87.964 | 97.036 | 3.915 | 62.526 | 137.785 | 14.345 | 42.308 | 26.670 | 3.887 |
| | | (9.713) | (169.446) | (0.195) | (1.412) | (122.810) | (1.749) | (1.685) | (10.214) | (0.301) |
| 3 | 30° Relax | 54.808 | 21.948 | 3.753 | 66.804 | 81.348 | 6.550 | 51.219 | 18.199 | 4.445 |
| | | (26.773) | (4.982) | (0.128) | (3.717) | (79.675) | (0.930) | (29.969) | (18.40) | (3.677) |
| | 35° Relax | 90.769 | 21.988 | 4.310 | 63.444 | 87.038 | 7.107 | 34.422 | 32.491 | 7.524 |
| | | (10.088) | (10.494) | (1.745) | (11.416) | (100.782) | (2.508) | (2.703) | (7.268) | (2.314) |
| | 30° Squat | 38.543 | 28.316 | 4.907 | 83.429 | 175.131 | 24.117 | 53.207 | 148.129 | 4.752 |
| | | (1.927) | (10.104) | (1.046) | (3.130) | (156.459) | (2.542) | (2.725) | (151.786) | (0.70) |
| | 35° Squat | 50.517 | 82.290 | 6.758 | 84.890 | 154.109 | 28.024 | 53.597 | 30.664 | 5.137 |
| | | (10.787) | (111.342) | (1.267) | (6.222) | (120.824) | (2.458) | (1.299) | (1.649) | (0.430) |
| 4 | 30° Relax | 73.626 | 79.669 | 3.948 | 61.752 | 55.481 | 10.113 | 31.995 | 34.694 | 11.289 |
| | | (26.850) | (97.517) | (0.519) | (12.070) | (23.814) | (4.178) | (2.537) | (10.227) | (1.287) |
| | 35° Relax | 86.538 | 29.208 | 3.220 | 61.457 | 143.718 | 10.712 | 33.489 | 38.640 | 12.103 |
| | | (9.993) | (1.739) | (0.148) | (6.237) | (138.906) | (3.499) | (2.80) | (9.845) | (1.992) |
| | 30° Squat | 124.118 | 19.641 | 2.925 | 55.110 | 50.702 | 20.341 | 44.643 | 89.829 | 5.515 |
| | | (56.30) | (8.320) | (1.114) | (2.093) | (6.083) | (1.741) | (8.871) | (125.361) | (0.620) |
| | 35° Squat | 100.733 | 31.535 | 6.086 | 74.118 | 51.423 | 27.023 | 79.462 | 86.004 | 6.876 |
| | | (18.913) | (12.979) | (1.599) | (1.973) | (6.480) | (4.445) | (19.203) | (111.911) | (2.036) |
| 5 | 30° Relax | 37.778 | 390.934 | 2.230 | 65.921 | 27.752 | 12.363 | 30.256 | 37.604 | 12.797 |
| | | (3.849) | (331.978) | (0.708) | (12.108) | (3.455) | (3.571) | (0.444) | (6.094) | (3.858) |
| | 35° Relax | 44.363 | 13.531 | 3.042 | 60.714 | 28.344 | 11.592 | 32.540 | 43.535 | 12.711 |
| | | (23.556) | (4.0) | (1.503) | (12.877) | (5.967) | (4.536) | (2.250) | (2.598) | (3.258) |
| | 30° Squat | 65.965 | 33.686 | 7.656 | 71.471 | 54.975 | 26.143 | 53.282 | 84.766 | 7.417 |
| | | (1.569) | (5.590) | (0.494) | (1.973) | (9.950) | (1.222) | (3.436) | (112.176) | (2.093) |
| | 35° Squat | 85.222 | 146.482 | 6.731 | 79.000 | 97.483 | 30.409 | 50.649 | 39.017 | 8.239 |
| | | (49.072) | (255.006) | (3.689) | (2.236) | (110.196) | (3.879) | (3.808) | (4.705) | (1.749) |
| 6 | 30° Relax | 166.667 | 160.377 | 1.688 | 25.029 | 50.760 | 16.263 | 63.158 | 32.994 | 11.876 |
| | | (125.831) | (156.309) | (0.587) | (1.044) | (12.735) | (1.50) | (0.0) | (4.244) | (0.973) |
| | 35° Relax | 85.812 | 132.362 | 2.794 | 25.370 | 59.799 | 14.678 | 63.529 | 27.592 | 8.834 |
| | | (98.911) | (198.677) | (1.093) | (0.811) | (0.561) | (2.123) | (6.113) | (3.635) | (1.418) |
| | 30° Squat | 112.933 | 16.018 | 2.113 | 73.118 | 44.175 | 3.370 | 75.000 | 58.672 | 15.759 |
| | | (79.810) | (6.616) | (0.567) | (8.318) | (10.518) | (0.328) | (0.0) | (17.394) | (2.312) |
| | 35° Squat | 71.784 | 31.395 | 2.903 | 100.104 | 43.089 | 2.574 | 78.095 | 92.083 | 21.413 |
| | | (5.737) | (15.574) | (0.257) | (14.923) | (56.386) | (0.191) | (10.433) | (106.052) | (3.447) |

Note: Freq unit is $s^{-1}$ and C unit is $s^{-1}$
Amp unit is maximum displacement of $m*s^{-2}$
Definitions of variables are in Appendix A.

**Table F-1. Continued**

| Subject | Condition | FreqX | CX | AmpX | FreqY | Cy | AmpY | FreqZ | CZ | AmpZ |
|---|---|---|---|---|---|---|---|---|---|---|
| 7 | 30° Relax | 65.476 | 19.543 | 3.657 | 53.987 | 77.053 | 10.151 | 29.536 | 34.681 | 7.685 |
| | | (35.053) | (1.637) | (0.196) | (16.528) | (90.085) | (1.893) | (1.123) | (1.428) | (0.970) |
| | 35° Relax | 69.571 | 22.948 | 3.785 | 54.762 | 29.991 | 7.581 | 69.524 | 140.082 | 6.400 |
| | | (19.434) | (5.088) | (0.579) | (4.124) | (4.571) | (1.70) | (69.698) | (153.809) | (1.226) |
| | 30° Squat | 49.936 | 76.389 | 5.362 | 75.235 | 221.954 | 13.852 | 50.175 | 48.033 | 3.805 |
| | | (8.856) | (24.136) | (1.045) | (4.708) | (197.739) | (4.282) | (3.318) | (14.745) | (0.70) |
| | 35° Squat | 63.714 | 169.882 | 5.244 | 72.478 | 45.570 | 9.072 | 47.320 | 57.013 | 2.080 |
| | | (18.615) | (182.402) | (0.402) | (16.733) | (16.172) | (0.80) | (2.106) | (97.844) | (0.203) |
| 8 | 30° Relax | 52.915 | 21.295 | 3.082 | 51.677 | 25.929 | 6.189 | 57.778 | 36.444 | 5.656 |
| | | (24.459) | (6.840) | (0.775) | (8.582) | (6.628) | (1.031) | (53.886) | (13.686) | (1.120) |
| | 35° Relax | 41.651 | 22.174 | 2.384 | 112.418 | 74.465 | 7.788 | 26.581 | 45.017 | 7.541 |
| | | (22.266) | (7.318) | (0.981) | (75.873) | (71.885) | (3.338) | (3.462) | (9.218) | (0.875) |
| | 30° Squat | 83.326 | 550.292 | 1.464 | 48.154 | 29.872 | 6.421 | 66.992 | 21.243 | 2.290 |
| | | (40.178) | (941.669) | (0.252) | (3.056) | (4.963) | (1.140) | (9.379) | (5.297) | (0.379) |
| | 35° Squat | 38.374 | 66.604 | 2.050 | 49.635 | 26.647 | 6.832 | 73.667 | 23.696 | 2.644 |
| | | (11.870) | (121.245) | (0.172) | (1.736) | (3.723) | (0.545) | (6.708) | (5.992) | (0.335) |
| 9 | 30° Relax | 27.775 | 189.703 | 1.322 | 57.492 | 44.695 | 2.840 | 28.034 | 192.686 | 3.989 |
| | | (7.716) | (196.128) | (0.297) | (5.50) | (17.056) | (0.642) | (2.282) | (138.264) | (0.658) |
| | 35° Relax | 23.817 | 14.501 | 1.664 | 63.492 | 204.295 | 3.602 | 28.826 | 82.710 | 5.584 |
| | | (2.133) | (8.421) | (0.148) | (5.499) | (271.343) | (0.603) | (1.070) | (99.309) | (0.708) |
| | 30° Squat | 58.654 | 18.143 | 2.829 | 65.673 | 104.367 | 9.316 | 76.822 | 26.926 | 2.817 |
| | | (11.740) | (6.350) | (0.432) | (12.832) | (121.626) | (1.306) | (18.810) | (9.551) | (0.345) |
| | 35° Squat | 74.496 | 22.318 | 2.624 | 93.429 | 87.965 | 7.404 | 80.462 | 19.567 | 2.637 |
| | | (6.556) | (5.998) | (0.368) | (33.527) | (89.224) | (1.333) | (7.078) | (10.480) | (0.195) |
| 10 | 30° Relax | 52.370 | 134.133 | 2.040 | 51.449 | 42.662 | 7.230 | 26.904 | 19.866 | 5.088 |
| | | (49.124) | (220.394) | (1.607) | (1.255) | (23.559) | (1.592) | (1.230) | (7.142) | (0.849) |
| | 35° Relax | 24.402 | 11.565 | 2.339 | 93.582 | 17.293 | 7.086 | 25.717 | 33.329 | 8.411 |
| | | (3.065) | (5.728) | (0.826) | (67.463) | (16.126) | (5.827) | (0.320) | (14.350) | (0.659) |
| | 30° Squat | 78.596 | 14.953 | 2.552 | 64.561 | 39.488 | 18.351 | 76.148 | 177.474 | 2.627 |
| | | (24.162) | (7.632) | (0.934) | (1.922) | (7.527) | (2.889) | (43.566) | (148.186) | (1.111) |
| | 35° Squat | 112.000 | 20.151 | 3.280 | 68.596 | 124.476 | 13.780 | 118.303 | 14.976 | 1.746 |
| | | (17.889) | (6.290) | (0.553) | (6.019) | (103.930) | (2.456) | (10.017) | (5.548) | (0.279) |
| 11 | 30° Relax | 80.238 | 17.515 | 3.645 | 54.386 | 30.389 | 12.203 | 81.905 | 91.740 | 7.639 |
| | | (5.361) | (1.282) | (0.486) | (10.427) | (7.182) | (0.579) | (3.299) | (99.302) | (0.607) |
| | 35° Relax | 71.515 | 20.164 | 2.746 | 60.759 | 20.552 | 8.596 | 72.719 | 27.160 | 6.268 |
| | | (14.696) | (12.737) | (0.267) | (8.611) | (5.477) | (2.438) | (8.650) | (4.247) | (0.802) |
| | 30° Squat | 64.561 | 83.909 | 3.508 | 57.600 | 47.919 | 16.625 | 78.880 | 102.252 | 4.109 |
| | | (1.922) | (131.980) | (0.416) | (5.367) | (7.898) | (0.980) | (9.461) | (108.342) | (0.592) |
| | 35° Squat | 67.584 | 30.488 | 4.844 | 67.027 | 36.273 | 17.643 | 82.604 | 20.733 | 4.238 |
| | | (12.654) | (12.577) | (0.282) | (9.287) | (5.837) | (3.532) | (6.618) | (4.513) | (0.476) |

Note: Freq unit is $s^{-1}$ and C unit is $s^{-1}$
Amp unit is maximum displacement of $m*s^{-2}$
Definitions of variables are in Appendix A.

## Appendix G

## Figures

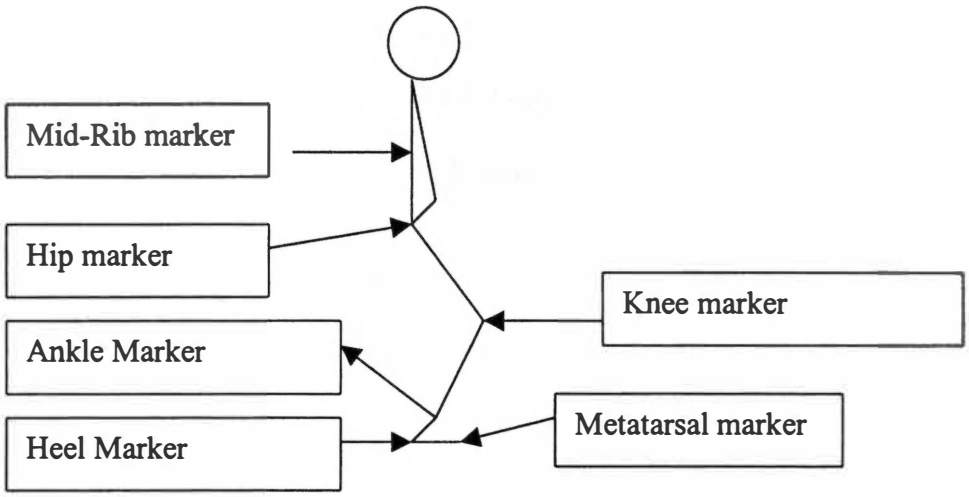



Figure G-1. Placement of reflective markers

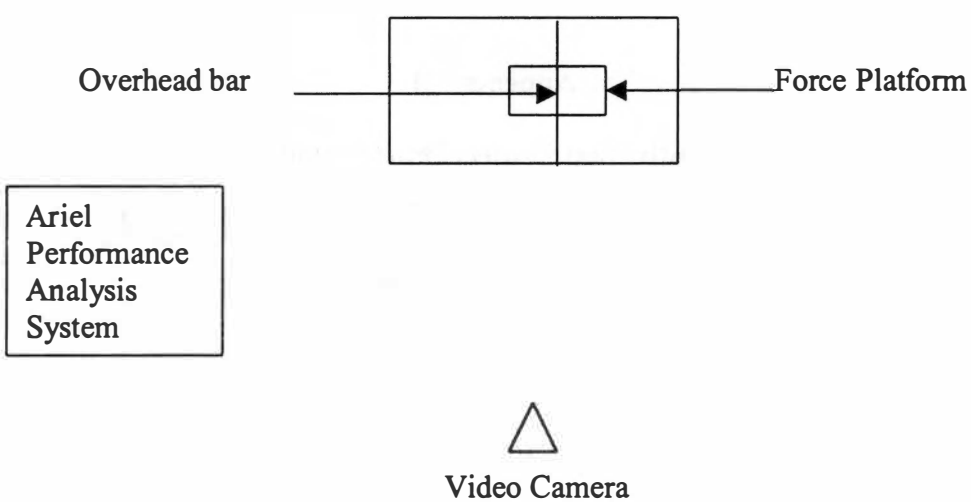



Figure G-2. Diagram of experimental setup

## Appendix H

## Individual Subject Information

**Table H-1. Individual Subject Information**

| Subject | Age | Body Mass | Height | Thigh Length | SKF |
|---|---|---|---|---|---|
| 2 | 22 | 71.81 | 172.72 | 40.0 | 8.33 |
| 3 | 23 | 69.40 | 180.34 | 41.0 | 7.33 |
| 4 | 22 | 92.30 | 180.34 | 42.0 | 10.00 |
| 5 | 25 | 90.50 | 182.88 | 45.1 | 9.67 |
| 6 | 24 | 89.12 | 172.72 | 39.7 | 14.00 |
| 7 | 23 | 79.78 | 177.80 | 42.6 | 10.33 |
| 8 | 21 | 93.15 | 180.98 | 45.5 | 13.33 |
| 9 | 20 | 80.58 | 187.96 | 46.9 | 7.67 |
| 10 | 18 | 76.23 | 187.96 | 44.5 | 5.20 |
| 11 | 22 | 85.52 | 185.42 | 45.5 | 9.67 |
| Mean | 22.0 | 82.84 | 180.91 | 43.28 | 9.55 |
| Std. | 2.0 | 8.57 | 5.46 | 2.55 | 2.66 |

Age is in years, Body mass is in kg.
Height is in cm, Thigh length is in cm.
SKF is in mm.

## Vita

Chuck Kohstall was in Columbus, OH on May 15, 1977. He attended Cedarville High School and graduated in 1995. He spent the 1995-1996 academic year as a freshman at Miami University (OH), and transferred the following academic year to The University of Dayton, OH and received a B.S. in Exercise Science/Fitness Management. In 2000, he began his graduate studies in Exercise Science with a concentration in Sports Medicine/Biomechanics at the University of Tennessee, Knoxville. Upon completion of his thesis, he received the Master of Science degree in Human Performance and Sport Studies in December of 2002.



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