AN ABSTRACT OF THE THESIS OF

Jeremy J. Bauer for the degree of Master of Science in Human Performance presented on April 27, 2000, Title: Kinetics and Kinematics of Prepubertal Children Participating in Osteogenic Physical Activity

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Abstract approved: ______________________________
Gerald A. Smith

Introduction: Recent reports in exercise related bone research have shown increased bone mineral content (BMC) at the femoral neck for prepubescent children participating in exercise programs consisting of repeated drop landings from a height of 61 cm. Increases in BMC from this type of exercise are believed to be the result of both high rate and magnitude of loading at the proximal femur. However, the dynamic characteristics associated with these landings in children have not been studied. **Purpose:** To describe the dynamic characteristics of children during landing and to quantify the forces associated with an activity associated with increases in bone mass. **Methods:** 13 prepubescent children (males=8, females=5, age 9.3 $\pm$ 0.7 years) who had previously completed drop landings over a 7 month period as part of an exercise intervention to increase bone mass participated in this research. Each subject performed 100 drop landings onto a force plate from a height of 61 cm. Ground reaction forces and two-dimensional kinematic data were recorded. Hip joint reaction forces were calculated using inverse dynamics based on a four segment rigid body model. Vertical ground reaction force and displacement data were fit to two single degree of freedom models, the Voigt and standard linear solid (SLS). The goodness of fit was quantified using the standard deviation of the error (SDE) between the experimental and the predicted data.

Results: Peak vertical ground reaction forces were 8.5 $\pm$ 2.2 (mean $\pm$ SD) body weights (BW) while hip joint reactions were 6.0 $\pm$ 1.8 BW. Loading rates for ground reaction forces during initial impact were in excess of 470 BW/s. Across 100 jump trials, ground reaction forces changed significantly for 5 subjects (4 increase, 1 decrease, $p < 0.05$) but were unchanged as a group. The SLS and Voigt models replicated the displacement traces well (SDE = 0.003 m and 0.001 m respectively). However, in fitting force data, the SLS outperformed the Voigt model (SDE = 580 N and 493 N respectively), but slightly under-predicted peak forces by 13%. **Conclusion:** Comparing force characteristics from drop landing to force characteristics known to be osteogenic, we can see how drop landings contribute to the osteogenic stimulus. The models used to represent children during drop landing closely fit displacement data, but did not replicate the time history of the impact force peaks thought to be important to osteogenesis. Quantification of exercises known to increase bone mass provides a basis on which to develop and implement additional exercise interventions for the purpose of increasing bone mass.

Kinetics and Kinematics of Prepubertal Children Participating in Osteogenic Physical Activity

By

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Dean of Graduate School

I understand that my thesis will become part of the permanent collection of Oregon State University libraries. My signature below authorizes release of my thesis to any reader upon request.

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Jeremy J. Bauer, Author

ACKNOWLEDGMENT

It takes a lot of people to accomplish great things. Therefore, please read about the people who have been instrumental in making this thesis happen. Before anything can be said about the research performed in this thesis I must thank the cool kids who let me put reflective tape all over them so they could jump 100 times in front of a camera and bright light onto a metal plate. While not the most exciting activity in the world, the implications of the research the children agreed to participate in could mean the difference between someone suffering serious debilitating fracture from a fall and falling without injury. The research is crucial to osteoporosis prevention.

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Stonick, served on my committee as a graduate school representative. I really appreciate her taking the time to participating in my education.

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CONTRIBUTION OF AUTHORS

Robyn Fuchs wrote most of the IRB forms from which our IRB was adapted. In addition, Robyn recruited all of the subjects for this study, assisted with data collection, and ran the intervention that made these children the topic of interest for this thesis. Dr. Gerald Smith assisted in study design, instrumentation, numerical methods for signal processing, inverse dynamics calculations and help in the editing of several drafts of this thesis. Dr. Christine Snow made the children from the bone research lab intervention available for use as subjects in this thesis, met weekly to discuss the progress of the study, provided thorough editing of each manuscript, and most importantly conceived of a majority of the study addressed in chapter 2. Dr. Mark Costello was invaluable in writing code, editing and conceptualizing the modeling work performed in chapter 3. Dr. Wilson C. Hayes was involved in conceptualizing and developing the ideas presented in chapter 3, improving on the original proposed research immeasurably by increasing the engineering aspects of the research.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

LIST OF SYMBOLS

k	Stiffness along interval t_i to t_{i+1}
$F(t)$	Force at time t
$x(t)$	Displacement at time t
W	Vertical ground reaction force.
k_V	Stiffness of spring in Voigt model.
k_U	Stiffness of series spring in standard linear solid model.
k_L	Stiffness of lower spring in standard linear solid model.
c_V	Damping in Voigt model.
c_L	Damping in standard linear solid.
y_V	Position of mass in Voigt model.
$\dot{y}_V$	Velocity of mass in Voigt model.
$\ddot{y}_V$	Acceleration of mass in Voigt model.
y_U	Position of mass in standard linear solid model.
$\ddot{y}_U$	Acceleration of mass in standard linear solid model.
y_L	Position of connection between series spring and lower spring/damper system in standard linear solid model.
$\dot{y}_L$	Velocity of connection between series spring and lower spring/damper system in standard linear solid model.
ω_n	Natural frequency.
ζ	Damping ratio.

DEDICATION

This thesis is dedicated to those people that give selflessly to advance the education and well being of others.

KINETICS AND KINEMATICS OF PREPUBERTAL CHILDREN PARTICIPATING IN OSTEOGENIC PHYSICAL ACTIVITY.

CHAPTER 1
INTRODUCTION

Identification of exercise programs that increase bone mass, particularly in children, is important to developing exercise prescriptions for osteoporosis prevention. To date, many exercise interventions have defined programs with little attention to the quantification of the force characteristics of the activities. From research performed using both animals and humans, we know that osteogenesis is related to the magnitude of applied forces and strain rate (Fehling, Alekel, Clasey, Rector, and Stillman, 1995; Lanyon and Rubin, 1984; Rubin & Lanyon, 1987). Identification and quantification of the force characteristics of the exercise activities required to increase bone mass is central to defining an exercise prescription for building bone.

Fuchs and Snow (1999) reported that prepubescent children performing 300 drop landings a week from a height of 61 cm over a 7-month period increased femoral neck bone mass 5.6% more than children in a control group. Characteristics of the forces resulting from drop landings are similar to those reported to be effective in increasing bone mass in animal models (Bauer et. al, 2000). Therefore, through analysis of exercise programs successful in increasing bone mass, researchers should discover force characteristics similar to those force characteristics related to increases in bone mass in carefully controlled animal studies (Rubin and Lanyon, 1987). In addition to assessing the force characteristics of an activity for comparison to in vitro animal studies, an overall shock response provided through dynamic modeling may provide the best indication of what the whole body experiences during impact activities.

The purpose of this thesis was to describe characteristics of drop landing by answering the following questions: 1) In prepubescent children what are the peak ground and hip joint reaction forces associated with drop landing from a height of 61 cm? 2) Do the peak reaction forces change across 100 trials? and 3) Can we use single degree of freedom dynamic models to describe the stiffness and damping properties of children during a drop landing event?

CHAPTER 2

KINETICS AND KINEMATICS OF PREPUBERTAL CHILDREN PARTICIPATING IN OSTEOGENIC EXERCISE

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2.1 Abstract

Drop landing exercise has been found to increase bone mass in children (Fuchs, R.K. & Snow, C.M., 1999). However, the forces associated with such exercises in children have not been studied. **Purpose:** To calculate the average peak ground and hip joint reaction forces associated with drop landing and to evaluate force changes across 100 trials. **Subjects:** 13 prepubescent children (males=8, females=5, age 9.3 ± 0.7 years) who had previously completed drop landings over a 7 month period as part of an exercise intervention to increase bone mass. **Methods:** Each subject performed 100 drop landings onto a force plate from a height of 61 cm. Ground reaction forces and two-dimensional kinematic data were recorded. Hip joint reaction forces were calculated using inverse dynamics based on a four segment rigid body model. Linear regression was used to assess changes in peak forces across trials within each subject and for the group. **Results:** Peak vertical ground reaction forces were 8.5 ± 2.2 (mean ± SD) body weights (BW) while hip joint reactions were 6.0 ± 1.8 BW. These force values are distributed between both legs. Loading rates for ground reaction forces were in excess of 470 BW/s. Across 100 jump trials, ground reaction forces changed significantly for 5 subjects (4 increase, 1 decrease, $p < 0.05$) but were unchanged as a group. **Conclusion:** Comparing force characteristics from drop landing to force characteristics known to be osteogenic, we can see how drop landings contribute to the osteogenic stimulus. Quantification of exercises known to increase bone mass provides a basis on which to develop and implement additional exercise interventions for the purpose of increasing bone mass.

2.2 Introduction

Participation in high force producing impact activities (i.e. those having ground reaction forces > 4 times body weight) is associated with increased bone mineral density at the hip (Fehling, Alekel, Clasey, Rector, and Stillman, 1995; Korht, Ehsani, and Birge, 1997; Taaffe, Robinson, Snow, and Marcus, 1997). Knowing the forces from activities associated with changes in bone mass provides researchers with the tools to develop exercise protocols for osteoporosis prevention. For example, gymnasts have greater bone mineral density at the hip than runners (Robinson et al., 1995). Comparing the two activities, gymnasts are subjected to greater ground reaction forces than runners and greater loading rates (Munro et al. 1997; McNitt-Gray, 1993; Panzer, 1987). These results have provided important information for designing bone loading exercise programs. Unfortunately, few studies have measured forces in exercise interventions designed to increase bone mass. In addition, had forces been measured it would be difficult to determine the individual contribution of each activity to the osteogenic response. If there is to be flexibility in the design of exercise programs for increasing bone mass we must first know the contribution of individual activities to osteogenesis and why they are osteogenic.

Using gymnasts as a model, Fuchs and Snow (1999) developed a highly specific exercise program designed to increase bone mass at the hip in prepubescent children (Tanner stage 1). Thirty-four boys and girls were randomly assigned to either a jumping (n=18) or control group (n=16). Jumpers performed drop landings from a height of 61 cm, 100 times in one 15-minute session, three times a week with an emphasis on landing as stiff as possible. To standardize contact time, the control group performed stretching activities. After 7-months the jumpers exhibited a 5.6% greater increase in bone mineral content at the femoral neck than the control group (Fuchs & Snow, 1999). Although a bone response was clear, it is not known what forces were associated with the change in bone nor whether the forces were constant across 100 trials. No studies to our knowledge have measured forces in children participating in exercise interventions designed to increase bone mass. In order to describe forces associated with specific changes in bone mass, a complete analysis of the force characteristics is required.

The purpose of this study was to describe the force characteristics of drop landing and answer the following questions: In prepubescent children what are the average maximum vertical ground and maximum resultant hip joint reaction forces associated with drop landing from a height of 61 cm? Do the maximum reaction forces change across 100 trials? By quantifying the drop landing exercise known to increase bone mass and comparing it to other activities such as running and walking we will provide a basis on which to develop and implement additional exercise interventions for the purpose of increasing bone mass.

2.3 Methods

Subjects. Thirteen prepubescent children (males=8, females=5, age 9.3 ± 0.7 years) who had previously completed drop landings over a 7 month period as part of an exercise intervention to increase bone mass participated in this research. All children were Tanner stage 1 as was assessed by line drawings of the tanner stages identified by the children and parents at the time of testing. This study was approved by the Oregon

State University Institutional Review Board. Parents and children provided written informed consent.

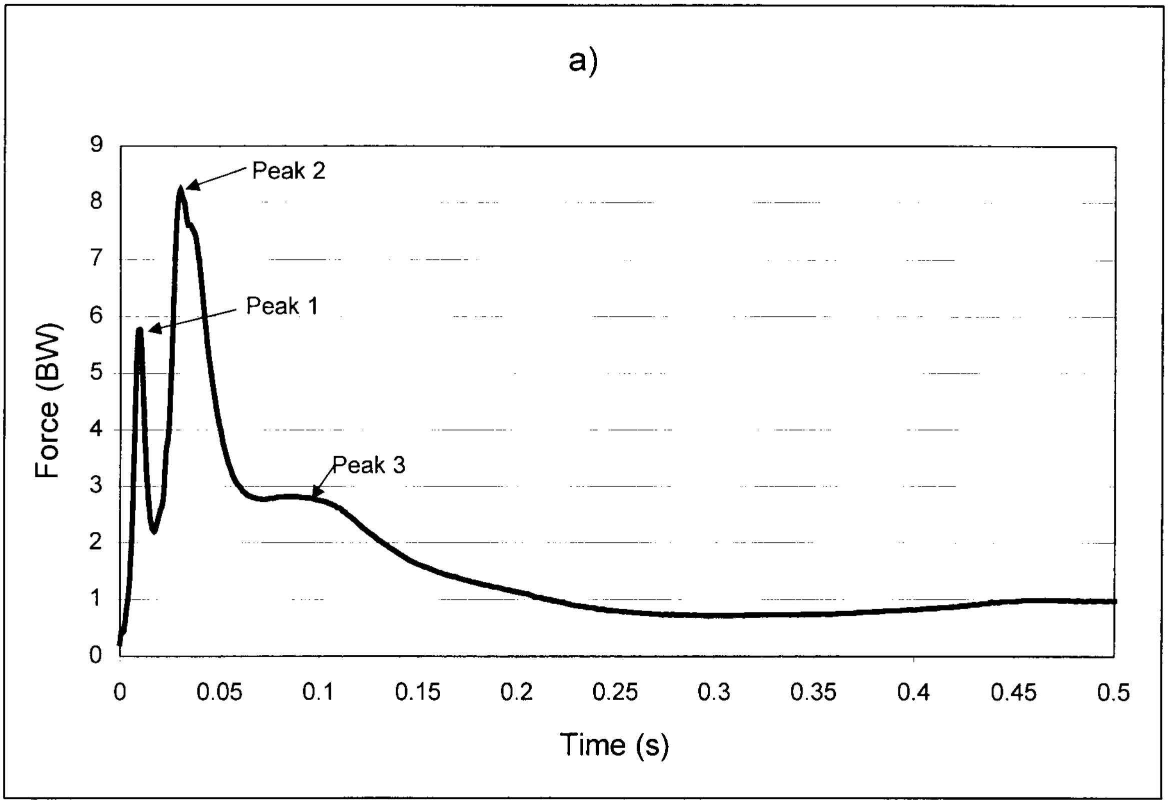
Task. Each subject performed 100 drop landings onto a force plate from a height of 61 cm in a period of approximately 15 minutes. After each landing, participants returned to the 61 cm height by first stepping onto a 30 cm high box, then to the 61 cm high box. Each subject was allowed to proceed through the 100 trials at his/her own pace.

Ground Reaction Forces. Subjects landed on a 0.60 x 0.40 m force platform (Kistler, 9281B) with both feet. Ground reaction forces were collected for 95 ms at 1000 Hz. Prior to each landing, an A/D board in a computer triggered collection of force data when a beam of light entering a photo resistor located 10 cm above the force plate was disrupted by each subject's feet. Triggering the force collection before contact provided data where the measured force should be zero. If the pre-contact data were nonzero then these values were subtracted from every force value in the respective trial to correct for a small amount of drift in the force plate transducers. A trial was "acceptable" when both feet were completely on the force platform from initial contact to standing at rest. Trials were excluded if the subject made contact with any surface other than the force platform upon landing and if the computer was not triggered to record ground reaction force data before initial contact. Only 80 of the 1300 trials were excluded.

Asymmetry between legs in the magnitude of ground reaction forces upon landing has been reported to be up to 14.8% (Schot, Bates and Dufek, 1994). An ideal assessment of the kinetics of each leg upon landing would require two force plates or landing on the force plate with one leg on and one leg off. However, for the purpose of this study it was assumed that each leg was subjected to exactly half of the total measured ground reaction force. It was felt the children would not proceed through the jumps naturally if they were required to target half of the force plate.

Loading rate, reported in body weights per second (BW/s), was determined by dividing the force at Peak 1 by the time to Peak 1 (Figure 2.1a)(Crossley et al., 1997). The common method used for calculating loading rate in running uses the portion of the force trace starting from 50 N up to 1 body weight + 50 N. While this is reasonable when using a ground reaction force trace from running, this method would neglect a large portion of the slope in the initial force peak from drop landing since the magnitude at the first peak is much higher than 1 body weight + 50 N.

Hip Joint Reaction Forces. Six 1-cm diameter reflective markers made from 3M retro-reflective tape were placed on the left side at the following anatomical sites: heel, 5th metatarsal head, lateral malleolus (ankle), knee joint center, greater trochanter (hip) and acromion process (shoulder) (Figure 2.2). The left side of the body was chosen because previous bone mass measurements were taken on the left proximal femur. Two-dimensional kinematic data from sagittal plane motion were collected at 250 Hz using a high-speed digital camera (Redlake Corporation, model 1000/s). To synchronize the kinematic data with the force data, a pulse was produced by a digital output from the A/D board at the instant the force plate was triggered. The pulse produced a white square in the upper left hand corner of the video image. To ensure there was no delay between triggering of the force plate and output of the synchronizing pulse to video, the output of the photo resistor on an oscilloscope screen was also recorded on the right hand side of the video image. Since each child was allowed to proceed at his/her own pace it was not possible to record every trial with the camera due to the nature of the recording system. An attempt was made to capture as many trials as possible for each subject. However, The number of trials recorded on video were not the same across subjects (range 11-22). Peak 5 motion analysis software (Peak Performance Technologies, Englewood, CO) was used to digitize and filter the digitized displacement data. Displacement data were filtered using a 4^{th} order Butterworth recursive digital filter at a cutoff frequency of 6 Hz to exclude high frequency noise resultant from the Peak 5 auto-digitizing process. This process involved a double pass, forward and backward, to cancel out any phase distortion. The optimal cutoff frequency for the filter was determined by the Peak 5 software to be 2 Hz using the Jackson Knee method, which optimizes the cutoff frequency of a signal by twice differentiating the residual between the raw signal and the filtered signal. However, it appeared as though a cutoff frequency of 2 Hz over smoothed the displacement data at impact. Since the displacement data were required for calculating segment velocity and acceleration values, a double filtering procedure was used which reduced the noise in the displacement data using a conservative cutoff, followed by additional filtering of the velocity data. Choosing a higher cutoff frequency, 6 Hz v. 2 Hz, for the displacement data prevented over smoothing of the impact. Despite the prefiltering at 6 Hz, differentiation magnified the remaining noise substantially. Therefore, to optimize for the acceleration data a 4^{th} order Butterworth recursive digital filter with optimal cutoff frequency between 23 and 28 Hz was used to remove high frequency noise from the velocity data. This optimal cutoff frequency was calculated using residual analysis and was calculated for each individual segment and direction (Winter, 1995).



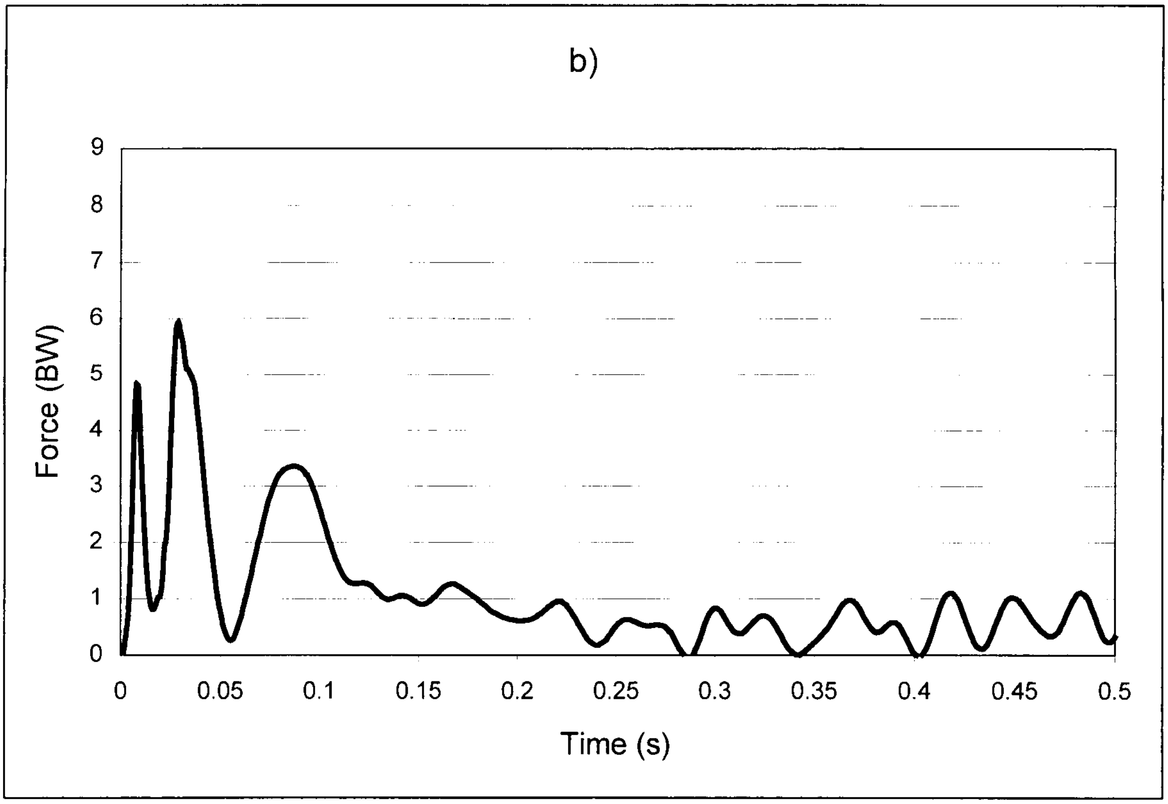


Figure 2.1. Typical traces from a) vertical ground reaction forces, and b) resultant hip joint reaction forces.

Figure 2.2. Jump sequence showing reflective marker placement.

The method of inverse dynamics used to calculate joint reaction forces is based on segment position and acceleration data which both need to be optimized to reduce any existing noise. Noise in the position data tends to occur from the digitizing process. Differentiation of position data into velocity and acceleration tends to magnify existing noise components substantially. Therefore, the general approach to the signal analysis was to filter twice. Filtering displacement data followed by filtering derivatives has been discussed extensively by Giakas & Baltzopoulos (1997).

The final step in the calculation of joint reaction forces involved merging the kinematic data with the force data. In order to use all of the 1000 Hz force data with the 250 Hz kinematic data, linear interpolation was used to determine kinematic data between samples, effectively creating a 1000 Hz kinematic data set. Body segment parameters specific to children were used to calculate segment percent of total body mass (Equations 2.1-3) and segment center-of-mass (Equations 2.4-6) (Table 2.1) (Jensen, 1986).

Mass proportions:

$$\begin{array}{ll} Foot & Y = 0.00015 \cdot age + 0.0187 \\ Shank & Y = 0.00122 \cdot age + 0.03809 \\ Thigh & Y = 0.00364 \cdot age + 0.06634 \end{array} \qquad \begin{array}{r} (2.1) \\ (2.2) \\ (2.3) \end{array}$$

Center of mass location:

$$\begin{array}{ll} Foot & Y = -0.00186 \cdot age + 0.4351 \\ Shank & Y = -0.003 \cdot age + 0.4526 \\ Thigh & Y = -0.00115 \cdot age + 0.4758 \end{array} \qquad \begin{array}{r} (2.4) \\ (2.5) \\ (2.6) \end{array}$$

The model used to calculate joint reaction forces is illustrated below using three rigid segments (foot, shank, thigh) (Figure 2.3). Prediction equations (Nigg & Herzog, 1995) for the hip reaction force components and resultant reaction force components were used (Equations 2.7-9).

Table 2.1. Body segment parameters.

		% of Total Mass			% to COM distance		
Individual	*Age*	*Foot*	*Shank*	*Thigh*	*Foot*	*Shank*	*Thigh*
Subj 1	9.2	2.0	4.9	10.0	41.8	42.5	46.5
Subj 2	8.4	2.0	4.8	9.7	41.9	42.7	46.6
Subj 3	8.8	2.0	4.9	9.8	41.9	42.6	46.6
Subj 4	9.6	2.0	5.0	10.1	41.7	42.4	46.5
Subj 5	8.5	2.0	4.8	9.7	41.9	42.7	46.6
Subj 6	9.0	2.0	4.9	9.9	41.8	42.6	46.5
Subj 7	10.0	2.0	5.0	10.3	41.7	42.3	46.4
Subj 8	8.4	2.0	4.8	9.7	41.9	42.7	46.6
Subj 9	9.9	2.0	5.0	10.2	41.7	42.3	46.4
Subj 10	10.1	2.0	5.0	10.3	41.6	42.2	46.4
Subj 11	8.8	2.0	4.9	9.8	41.9	42.6	46.6
Subj 12	10.2	2.0	5.1	10.3	41.6	42.2	46.4
Subj 13	9.7	2.0	5.0	10.2	41.7	42.4	46.5
Average	9.3	2.0	4.9	10.0	41.8	42.5	46.5

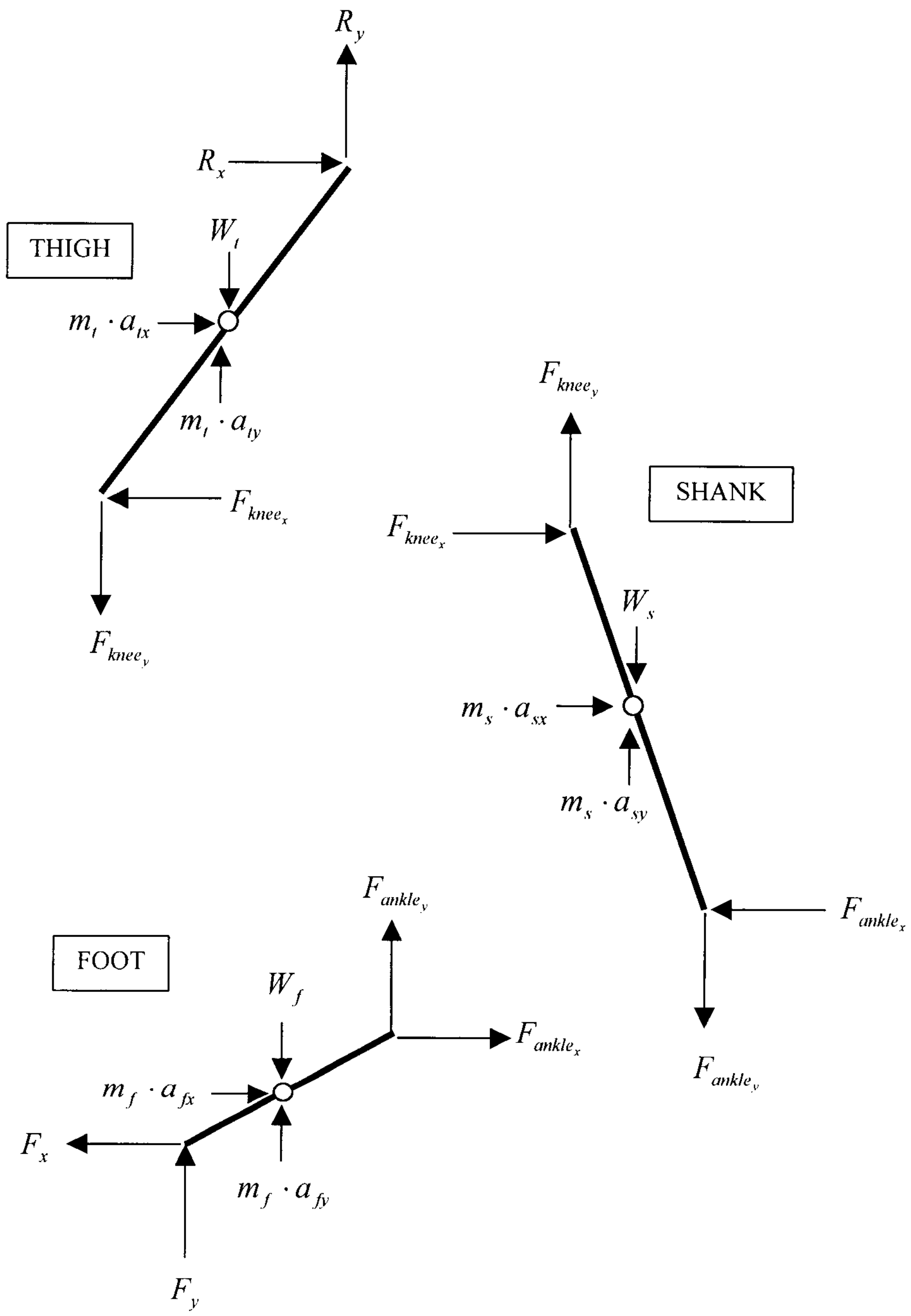


Figure 2.3. Joint reaction force free body diagrams

$$R_x = -F_x + m_f \cdot a_{fx} + m_s \cdot a_{sx} + m_t \cdot a_{tx} \tag{2.7}$$

$$R_y = W_f + W_s + W_t - F_y + m_f \cdot a_{fy} + m_s \cdot a_{sy} + m_t \cdot a_{ty} \tag{2.8}$$

$$R = \sqrt{R_x^2 + R_y^2} \tag{2.9}$$

where:

a_f	Acceleration of the foot.
a_s	Acceleration of the shank.
a_t	Acceleration of the thigh.
m_f	Mass of the foot.
m_s	Mass of the shank.
m_t	Mass of the thigh.
F_x	Horizontal ground reaction force.
F_y	Vertical ground reaction force.
R_x	Horizontal hip joint reaction force.
R_y	Vertical hip joint reaction force.
R	Resultant hip joint reaction force.
W_f	Weight of the foot.
W_s	Weight of the shank.
W_t	Weight of the thigh.

Statistical Analysis. Changes in peak vertical ground and hip joint reaction forces within each subject and for the group across 100 trials were analyzed using linear regression. The slope of the regression line for each subject and for the group was used to determine whether forces changed across trials (i.e. if slope was not significant different from 0 then peak ground reaction forces did not change across 100 trials). Analysis of variance was used to determine if there were any gender differences in ground reaction forces. Statistical significance was set at $p = 0.05$.

2.4 Results

Vertical Ground Reaction Force Trace. The typical force trace observed during landing has three distinguished peaks consistent with measurements from other landing literature (Figure 2.1a) (Devita & Skelly, 1992; Dufek & Bates, 1990). The first peak represents toe contact, the second peak, with generally the greatest magnitude, is heel contact and the third peak is representative of active muscle activation slowing the descent of the center of mass. While no EMG data were recorded during the landings in this study, sharp peaks in a force trace have been explained as a result of passive reflexive muscle stiffness (Dyhre-Poulsen, Simonsen, & Voigt, 1991). An active peak force, due to voluntary muscular contraction in braking movements, usually follows the initial passive peaks.

Ground Reaction Forces. The force at Peak 1 was 5.6 ± 1.4 (mean ± SD) times body weight (BW) (Table 2.2). Time to Peak 1 was 0.012 ± 0.003 s. The force at Peak 2 was 8.5 ± 2.3 BW. Time to Peak 2 was 0.038 ± 0.006 s. Boys landed significantly harder than girls (9.3 ± 2.4 BW v. 7.4 ± 1.5 BW, $p < 0.01$). Loading rate, calculated using only Peak 1 and the time to Peak 1was 472 ± 168 BW/s.

Five of the thirteen subjects had statistically significant (non-zero peak GRF regression line slopes) changes in maximum ground reaction forces across 100 trials

(Figure 2.4, Table 2.2). Four of the subjects, with changes across trials, increased maximum ground reaction forces as trial number increased ($p < 0.001$) and one subject decreased maximum ground reaction forces as trial number increased ($p < 0.02$). As a group, maximum ground reaction forces did not change across trials ($p = 0.11$).

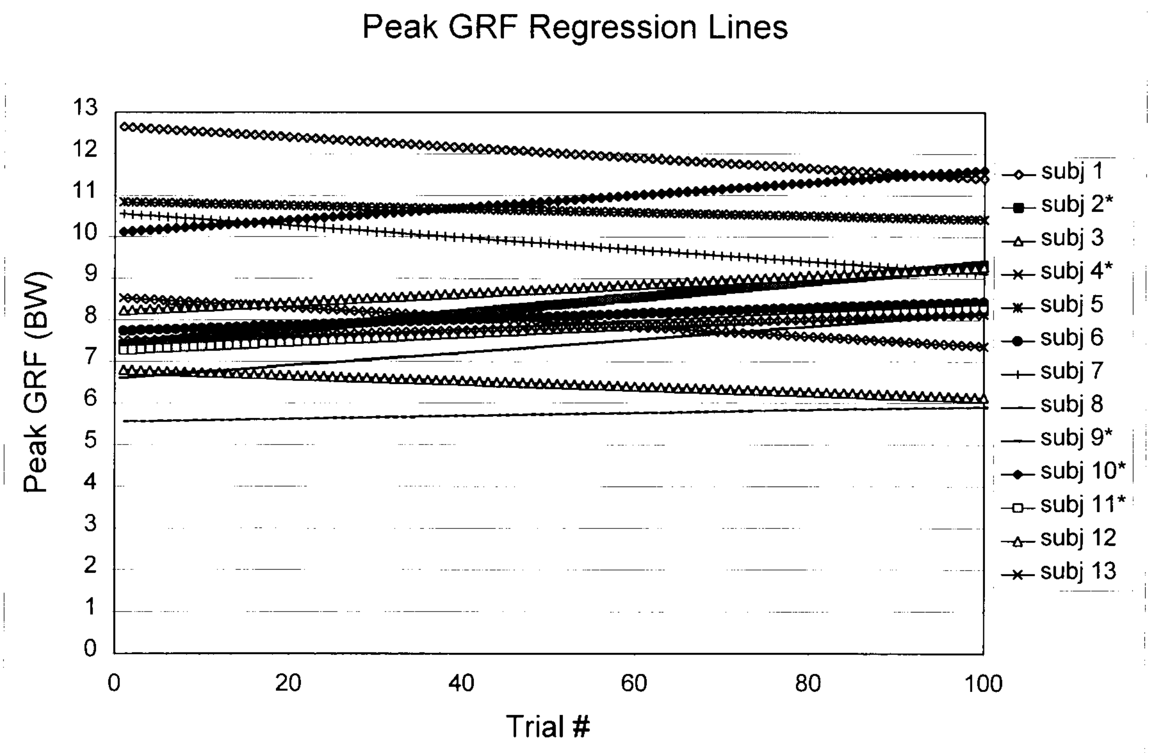


Figure 2.4. Ground reaction force trend lines (* $p < 0.05$).

Hip Joint Reaction Forces. The average maximum resultant hip joint reaction force for all subjects was 6.0 ± 1.8 BW (Figure 2.1b) (Table 2.2). Maximum resultant hip joint reaction forces did not change significantly across trials for any subject or for the group.

Table 2.2. Summary results

Individual	*Height (cm)*	*Mass (kg)*	*Age (y)*	*Gender*		*Mean*	*SD*	*n*	*Slope*
Subj 1	131.5	29.1	9.2	M	*yGRF (BW)*	12.0	2.0	99	-0.013
					rHJRF (BW)	9.5	2.1	19	-0.033
Subj 2	121.9	27.5	8.4	F	*yGRF (BW)*	8.4	1.2	100	*0.0194
					rHJRF (BW)	6.7	0.9	20	-0.013
Subj 3	132.7	26.4	8.8	M	*yGRF (BW)*	6.5	1.6	92	-0.007
					rHJRF (BW)	5.1	1.4	17	0.015
Subj 4	139.7	30.0	9.6	F	*yGRF (BW)*	7.9	1.5	98	*-0.0118
					rHJRF (BW)	6.3	1.2	19	-0.003
Subj 5	130.8	26.6	8.5	M	*yGRF (BW)*	10.6	1.9	71	-0.004
					rHJRF (BW)	8.2	1.6	11	-0.005
Subj 6	142.2	35.8	9.0	M	*yGRF (BW)*	8.1	1.1	100	0.007
					rHJRF (BW)	5.3	0.8	20	-0.001
Subj 7	139.1	30.9	10.0	M	*yGRF (BW)*	9.8	1.8	70	-0.015
					rHJRF (BW)	6.4	0.6	17	-0.002
Subj 8	140.3	33.5	8.4	F	*yGRF (BW)*	5.7	0.9	99	0.004
					rHJRF (BW)	4.1	0.6	19	0.008
Subj 9	140.3	44.3	9.9	F	*yGRF (BW)*	7.4	1.2	98	*0.0156
					rHJRF (BW)	6.0	0.6	18	0.000
Subj 10	144.2	47.7	10.1	M	*yGRF (BW)*	10.9	1.3	99	*0.0149
					rHJRF (BW)	6.1	1.0	22	0.012
Subj 11	140.3	34.9	8.8	F	*yGRF (BW)*	7.8	0.9	100	*0.0101
					rHJRF (BW)	4.4	0.7	15	-0.003
Subj 12	143.5	53.7	10.2	M	*yGRF (BW)*	8.8	1.6	100	0.011
					rHJRF (BW)	5.2	1.3	20	-0.001
Subj 13	141.0	40.1	9.7	M	*yGRF (BW)*	7.8	1.4	94	0.006
					rHJRF (BW)	5.1	0.7	18	0.000
Average	137.5	35.4	9.3		*yGRF (BW)*	8.5	2.2	1220	0.003
					rHJRF (BW)	6.0	1.8	235	-0.002

yGRF = Vertical ground reaction force
rHJRF = Resultant hip joint reaction force
*Statistically significant ($p<0.05$) nonzero regression line slopes.

2.5 Discussion

In this study we sought to describe drop landing and answer two specific questions. First, what are the force characteristics associated with drop landing? Upon landing, children were subjected to two sharp force peaks visible in both the ground reaction force profiles and the hip joint reaction force profiles. In the ground reaction force profiles, one peak was in excess of 5 BW and the other in excess of 8 BW. The average maximum hip joint reaction force was around 6 BW. Again, it should be pointed out that the force values reported here are for both legs.

In the second question we asked, do the forces change across 100 trials? The children in this study easily completed 100 trials in a session and maintained relatively constant force characteristics as a group. Five subjects had statistically significant changes in maximum vertical ground reaction forces across 100 trials. While the slopes of the regression lines were extremely small, it can not be concluded that all of the children were subjected to consistent forces across all 100 trials (Table 2.2). However, with nearly 100 trials in each subject's data set, slight changes of force across trials were likely statistically detectable beyond levels of practical significance. While the slope of a regression line may represent a decrease of 1 BW from trial 1 to trial 100, that 1 BW difference may not significantly alter the osteogenic stimulus. Therefore, the slope needed to represent a practical change in forces across trials is not known at this time.

One of the strengths to this study was that the subjects used were from a population of children that had significantly increased bone mineral content at the femoral neck as a direct result of performing repeated drop landings. Each subject was familiar with the task, did not require instruction, and was allowed to proceed through the jumps at his/her own pace. While some trials had to be excluded after data collection, ground reaction forces were recorded for all 100 trials for every subject. In addition, high sampling rates were used for collecting both kinematic and force data.

Several limitations in the study reduced our ability to provide a complete analysis of the drop landing activity in children. First, the subjects had not performed the drop landing exercises for 6-months prior to participating in this study. By detraining for 6-months, normal growth in the children could have caused coordination changes from the original exercise intervention. However, since forces did not change substantially over 100 trials, we believe that the forces collected after detraining accurately reflect those during training. Another limitation was that joint moments were not calculated. Although moment calculations would have provided important information concerning net muscle forces acting at the hip in addition to the joint reaction forces calculated using rigid body dynamics, landing with two feet on one force plate made it difficult to use the recorded center of pressure information needed for 2-D moment calculations on one leg. Two footed landings were required for mechanical modeling research that will be reported outside of this manuscript. In addition, it was felt that targeting half of the force plate would change the landing from what was performed during the exercise intervention. Moments could have been calculated using the 250 Hz kinematic data working from the top down to the hip. However, working from the top down would have required full body kinematics and would likely involve somewhat larger measurement uncertainties than working from the ground reaction force ground up procedure.

Magnitude, rate of loading and frequency of loading are factors that have been investigated for osteogenesis (Lanyon & Rubin, 1984; Mosley, & Lanyon, 1998;

O'Connor, & Lanyon, 1982; Rubin & Lanyon, 1987; Turner, Owan, & Takano, 1995; Whalen, Carter & Steele, 1988). Theoretically the drop landing exercise used in our study appears to provide the bone with a more effective osteogenic stimulus than is possible with either running or walking due to the nature of the loading. Maximum ground reaction forces measured in adults during walking range from 1 - 2 BW with a maximum force occurring after about 0.1s (Crossley et al., 1997). Maximum ground reaction forces measured in adults during running range from 2 -3 BW with an impact peak from heel strike occurring after about 0.1 s (Breit and Whalen, 1997). Magnitudes of loading resulting from landing from a height of 61 cm are in excess of 8.5 times body weight at the ground and 6 times body weight at the hip for both legs. To more appropriately compare the reported drop landing ground reaction forces to ground reaction forces recorded during running and walking the forces from drop landing would need to be divided in half for each leg assuming perfect symmetry between legs giving values greater than 4 times BW for each leg during drop landing. Rubin and Lanyon (1987) reported that greater magnitude forces were more osteogenic then low magnitude forces. While the ground reaction forces from drop landing for each leg are only about 1 body weight greater than ground reaction forces from running, the novel act of repeated drop landings in a daily routine may be the reason for the large increases in bone mass in active children. Adding the drop landing stimulus to the daily routine of children who already walk and run throughout a day would provide the bone with loading it is not as familiar with, potentially causing the bone to change more than if the common activity of running was used as the intervention.

Mosely and Lanyon (1998) reported that fast strain rates are more osteogenic than slow strain rates. Loading rates while walking at 1.2 m/s have been reported to be 43 BW/s (Crossley et al., 1997). Loading rates while running at 4.3 m/s have been reported to be 63 BW/s. Munro et al. (1987) report loading rates in running at speeds ranging from 3 – 5 m/s to be 77 – 113 BW/s respectively. We report loading rates in excess of 470 BW/s for two footed landings. Dividing this value by two to address each leg individually, the rate of force production is more than two times the rate reported in either walking or running. However, we must be careful when comparing loading rates to strain rates. Loading rate is strictly the rate which a load is applied, whereas strain rate is the rate at which a structure, in this case bone, is deformed. Due to the viscoelastic nature of bone a stress strain diagram for bone loaded at a fast rate would have a steeper slope in the elastic region compared to a stress strain diagram for bone loaded at a slow rate. Therefore, a fast rate of loading could result in a lower magnitude strain compared to a slow rate of loading. It should not be concluded that fast rates of loading from drop landing are necessarily more osteogenic than the slower rates of loading reported in running and walking. However, such an extreme difference in loading rates between drop landing and walking and running raises some questions on how the loading rate actually relates to the strain rate and subsequently the osteogenic response as seen in animal studies.

While running and walking allow for a large amount of loading cycles, Rubin and Lanyon (1987) and Whalen, Carter and Steele (1988) have reported that the number of times a load is applied is not as important as the magnitude of the applied load. In fact, when loading turkey ulnas, Rubin and Lanyon found no differences in osteogenic activity between loading the ulnas 36 times a day compared with 1800 times a day. All of the children in our study, when asked to land as they did in their earlier exercise program, chose to land in a toe-heel fashion (Fuchs, & Snow, 1999). Initial contact was always with the toe followed by the heel producing two high magnitude force peaks (Figure 2.1a). The two ground reaction force peaks also translated to two joint reaction force peaks at the hip (Figure 2.1b). This translates to a total of 200 high magnitude loads per exercise session. Whether or not fewer cycles at the same magnitude would provide a similar bone response is unclear, but worth investigating.

The hip joint reaction forces were calculated using a simple rigid body model and should only be interpreted as the forces resulting from force transduction of the ground reaction forces through the lower skeleton. No attempt has been made to assess joint forces resultant from muscular contraction during the landings. We chose to use the rigid body model for hip joint reaction force estimations because it likely underestimated the real joint reaction forces compared to other methods such as accelerometry, the wobbling mass model or models that include estimated muscle forces, (Bogert, Read, and Nigg, 1996; Bogert, Read, and Nigg, 1999; Gruber et al., 1998; Röhrle et al., 1984). These less conservative estimating methods would less clearly provide a lower bound of likely force magnitudes.

An estimate of hip joint reaction forces resultant from an activity that has been associated with increases in bone mass in children is very important for exercise prescription. Therefore, studying and comparing the force characteristics at the site of interest with the easily measured ground reaction forces gives those in the bone research field a much better idea of how ground reaction forces relate to joint reaction forces at clinically relevant sites for bone measurement. The seven months of jump exercise with the force characteristics reported in this paper resulted in increases of hip bone mass of 5.6% more in jumpers than controls. While the results from our current research can only be generalized to children that have participated in the exercise program of drop landings, a great deal of information has been gathered concerning how children respond to this type of activity. Our results provide a quantitative basis from which to pursue exercise for bone mass accretion in children. Research investigating the relationship between exercise and bone mass should quantify the loading rate and the ground reaction forces resultant from participating in the exercise. Consistently assessing forces and loading rates in activities investigated for osteogenic effects will provide the field with a more clear picture of what characteristics result in osteogenesis. Quantification of exercises known to increase bone mass provides a basis on which to develop and implement additional exercise interventions for the purpose of increasing bone mass.

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CHAPTER 3

SINGLE DEGREE OF FREEDOM MODELS REPRESENT CHILDREN PARTICIPATING IN OSTEOGENIC EXERCISE

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3.1 Abstract

Recent reports have shown increased bone mineral content at the femoral neck for prepubescent children participating in drop landings. The bone response from this exercise is believed to result from the high force magnitude and fast loading rates to which the proximal femur is subjected. Typically, the use of simple dynamic models to characterize the applied forces at the ground with center of mass displacement has neglected the aspects of the force signal thought to be crucial to osteogenesis. To better understand the dynamic characteristics of a drop landing, our aim was to answer the following questions: 1) Can single degree of freedom models represent ground reaction force and center of mass displacement characteristics of children drop landing? and 2) What are the stiffness and damping properties of children participating in drop landing? Ground reaction forces were recorded for 13 prepubescent children who completed 100 drop landings from a height of 61 cm. Vertical ground reaction force and displacement data were fit to two single degree of freedom models, the Voigt and standard linear solid (SLS). The goodness of fit was quantified by calculating the standard deviation of the error (SDE) between the experimental data and the predicted data. The SLS and Voigt models replicated the displacement traces well (SDE = 0.003 m and 0.001 m respectively). However, in fitting force data, the SLS outperformed the Voigt model (SDE = 580 N and 493 N respectively), but slightly under-predicted peak forces by 13%. Single degree of freedom models closely fit displacement data, but did not replicate impact force peaks thought to be important to osteogenesis.

3.2 Introduction

Identification of exercise programs that increase bone mass, particularly in children, is important to developing exercise prescriptions for osteoporosis prevention. To date, many exercise interventions have defined programs with little attention to the quantification of the force characteristics of the activities. From research performed on both animals and humans, we know that osteogenesis is dependent on the magnitude of applied forces and strain rate (Fehling, Alekel, Clasey, Rector, and Stillman, 1995; Rubin & Lanyon, 1987). Identification and quantification of the kinetic characteristics of the exercise activities required to increase bone mass is central to defining an exercise prescription for building bone.

Fuchs and Snow (1999) reported that prepubescent children performing 300 drop landings a week from a height of 61 cm over a 7-month period increased femoral neck bone mass 5.6% more than children in a control group. Characteristics of the forces resulting from drop landings are similar to those reported to be effective in increasing bone mass in animal models (Bauer et. al, 2000). Therefore, through analysis of exercise programs successful in increasing bone mass, researchers should discover force characteristics similar to those force characteristics related to increases in bone mass in carefully controlled animal studies (Rubin and Lanyon, 1987). In addition to assessing the force characteristics of an activity for comparison to in-vitro animal studies, an overall shock response represented by dynamic modeling may provide the best indication of what the whole body experiences during impact activities.

Dynamic simulation of gross motion of the lower extremity has generally involved relatively simple models. For example, the mass-spring model has been used to describe stiffness of the leg and total body in activities such as hopping, running and drop landing (Blickhan, 1989; Ferris, Louie and Farley, 1998; Farley et al., 1998; Farley et al., 1991; Ferris and Farley, 1997; McMahon and Cheng, 1990; Farley and Gonzelez, 1996; Dalleau et al., 1998; Schepens, Willems and Cavagna, 1998). Schepens, Willems and Cavagna (1998) plotted vertical acceleration versus vertical displacement of the center of mass using force data collected from children running across a force plate. Stiffness was calculated as the slope of the portion of the trace containing only upward displacement of the center of mass, neglecting the acceleration data recorded at initial impact which contained force peaks resultant from heel strike. Farley and Gonzalez (1996) suggested that the relationship between force and displacement of the center of mass while running was linear *following* the sharp force peak at impact from heel strike. Ferris and Farley (1997) calculated vertical stiffness during hopping as the ratio of the maximum vertical ground reaction force to the maximum vertical displacement of the center of mass, eliminating all other forces from the stiffness calculation. Neglecting impact peaks in the modeling process eliminates key characteristics of the force trace that are thought to be important for osteogenesis. However, it is easy to see how the initial impact peaks from from running can skew a mostly linear trace of ground reaction force plotted versus center of mass displacement (Figure 3.1, 3.2).

Drop landing ground reaction force traces are characterized by two initial peaks representing initial contact with the forefoot followed by heel contact (Figure 3.3). The

Figure 3.1. Force and displacement traces from running.

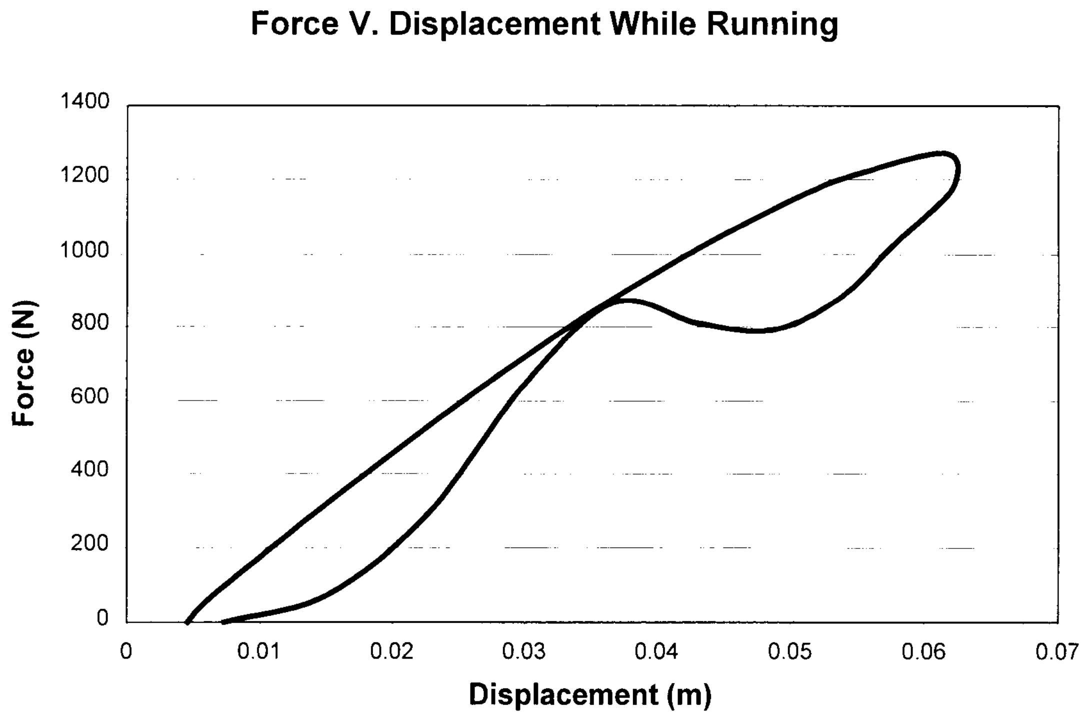


Figure 3.2. Force versus displacement graph used for estimating vertical stiffness.

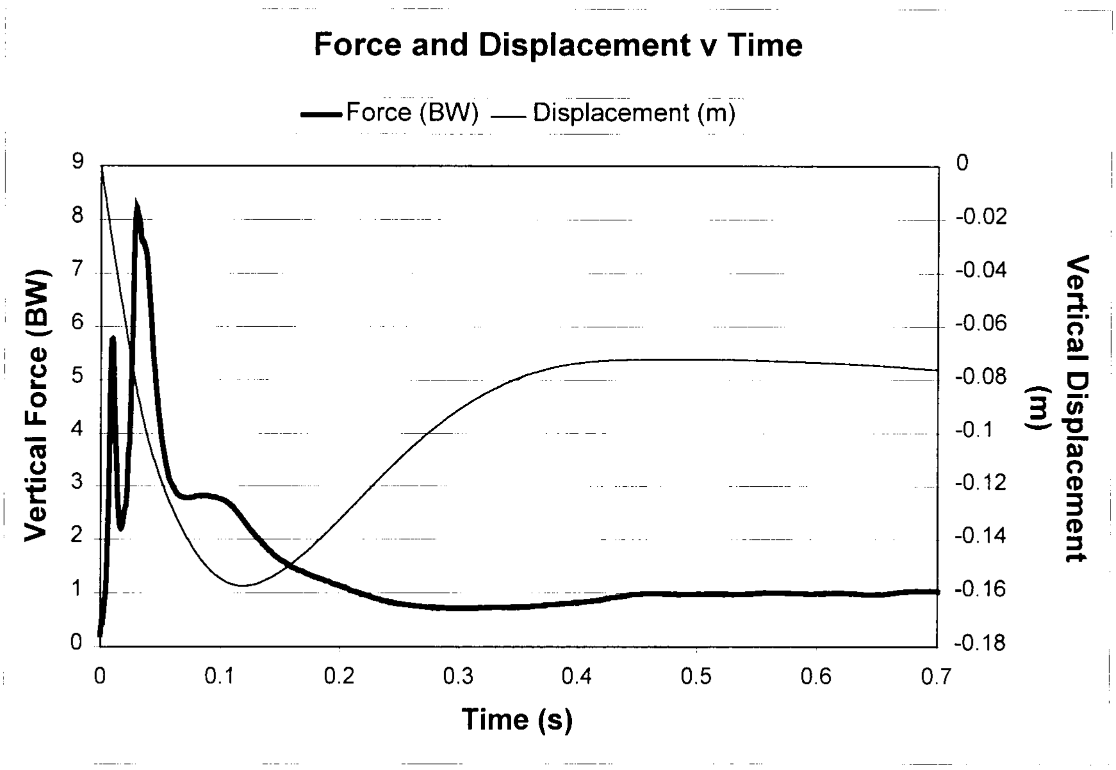


Figure 3.3. Typical force v. center of mass displacement from drop landing.

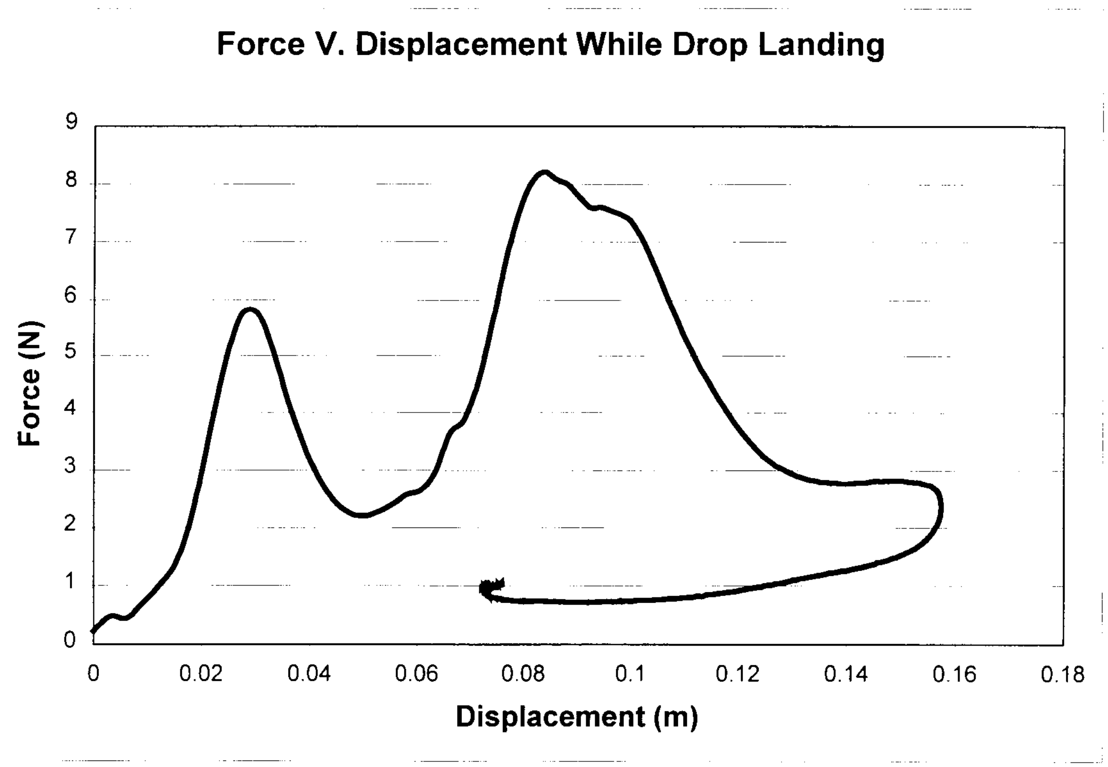


Figure 3.4. Force versus displacement graph for estimating vertical stiffness.

two impact peaks are followed by a third peak that is substantially lower in magnitude. The displacement trace shows a single peak, which occurs around the same time as the third force peak. While single degree of freedom mass spring models have been successfully employed for repetitive activities where the mass center oscillates sinusoidally (i.e. running and hopping), this is not the case for drop landings where the displacement data more closely resemble a damped sinusoid. In addition, the displacement data represent a global response to the entire drop landing event whereas the commonly neglected impact forces need to be addressed as well. When trying to calculate vertical stiffness during drop landing by plotting force versus center of mass displacement, one can see that the impact peaks prevent any reasonable value from being obtained because the curve is non-linear throughout (Figure 3.4). As we gain a better understanding of those aspects of the time history of the ground reaction force trace that associate most strongly with the osteogenic response, we could modify the simple models mentioned above to simulate the ground reaction force and displacement time history of a drop landing.

The purpose of this research was to answer the following questions: 1) Can single degree of freedom models represent ground reaction force and center of mass displacement characteristics of children during drop landing? and 2) What are the stiffness and damping properties of children participating in drop landing activity?

3.3 Methods

Subjects. Thirteen prepubescent children (males=8, females=5) age 9.3 ± 0.7 years (Table 3.1) that had increased bone mass as a result of completing 7-months of drop landings participated in this study. Parents and children provided written informed consent to participate. The study was approved by the Oregon State University Institutional Review Board.

Task. Each subject performed 100 drop landings (Figure 3.5) from a height of 61 cm in a period of approximately 15 minutes. After every landing each subject returned to the 61 cm height by first stepping onto a 30 cm high box then to the 61 cm box. Each subject was allowed to proceed through the 100 trials at his/her own pace.

Figure 3.5. Drop landing event.

Table 3.1. Maximum force and displacement summary data.

Subject	Height (cm)	Mass (kg)	Age (y)	Gender		Mean ± SD	n
Subj 1	131.5	29.1	9.2	M	yGRF (BW)	12.0 ± 2.0	99
					pGRF (BW)	11.1 ± 1.7	99
					DISP (m)	0.19 ± 0.03	99
Subj 2	121.9	27.5	8.4	F	yGRF (BW)	8.4 ± 1.2	100
					pGRF (BW)	7.4 ± 1.2	100
					DISP (m)	0.18 ± 0.02	100
Subj 3	132.7	26.4	8.8	M	yGRF (BW)	6.5 ± 1.6	92
					pGRF (BW)	5.6 ± 1.5	92
					DISP (m)	0.30 ± 0.11	92
Subj 4	139.7	30.0	9.6	F	yGRF (BW)	7.9 ± 1.5	98
					pGRF (BW)	7.7 ± 2.2	95
					DISP (m)	0.26 ± 0.06	98
Subj 5	130.8	26.6	8.5	M	yGRF (BW)	10.6 ± 1.9	71
					pGRF (BW)	9.4 ± 1.8	69
					DISP (m)	0.25 ± 0.09	71
Subj 6	142.2	35.8	9.0	M	yGRF (BW)	8.1 ± 1.1	100
					pGRF (BW)	7.1 ± 1.0	100
					DISP (m)	0.18 ± 0.02	100
Subj 7	139.1	30.9	10.0	M	yGRF (BW)	9.8 ± 1.8	70
					pGRF (BW)	8.8 ± 2.0	68
					DISP (m)	0.22 ± 0.06	70
Subj 8	140.3	33.5	8.4	F	yGRF (BW)	5.7 ± 0.9	99
					pGRF (BW)	4.8 ± 0.9	99
					DISP (m)	0.30 ± 0.06	99
Subj 9	140.3	44.3	9.9	F	yGRF (BW)	7.4 ± 1.2	98
					pGRF (BW)	6.6 ± 1.2	98
					DISP (m)	0.21 ± 0.03	98
Subj 10	144.2	47.7	10.1	M	yGRF (BW)	10.9 ± 1.3	99
					pGRF (BW)	9.1 ± 1.3	99
					DISP (m)	0.19 ± 0.03	99
Subj 11	140.3	34.9	8.8	F	yGRF (BW)	7.8 ± 0.9	100
					pGRF (BW)	6.6 ± 0.8	100
					DISP (m)	0.20 ± 0.03	100
Subj 12	143.5	53.7	10.2	M	yGRF (BW)	8.8 ± 1.6	100
					pGRF (BW)	8.0 ± 1.5	100
					DISP (m)	0.19 ± 0.04	100
Subj 13	141.0	40.1	9.7	M	yGRF (BW)	7.8 ± 1.4	94
					pGRF (BW)	7.1 ± 1.3	90
					DISP (m)	0.29 ± 0.09	94
Total	137.5	35.4	9.3		yGRF (BW)	8.5 ± 2.2	1220
					pGRF (BW)	7.6 ± 2.2	1209
					DISP (m)	0.23 ± 0.07	1220

yGRF = experimental peak vertical ground reaction force.
pGRF = predicted peak ground reaction force using the standard linear solid.
DISP = experimental peak center of mass displacement.

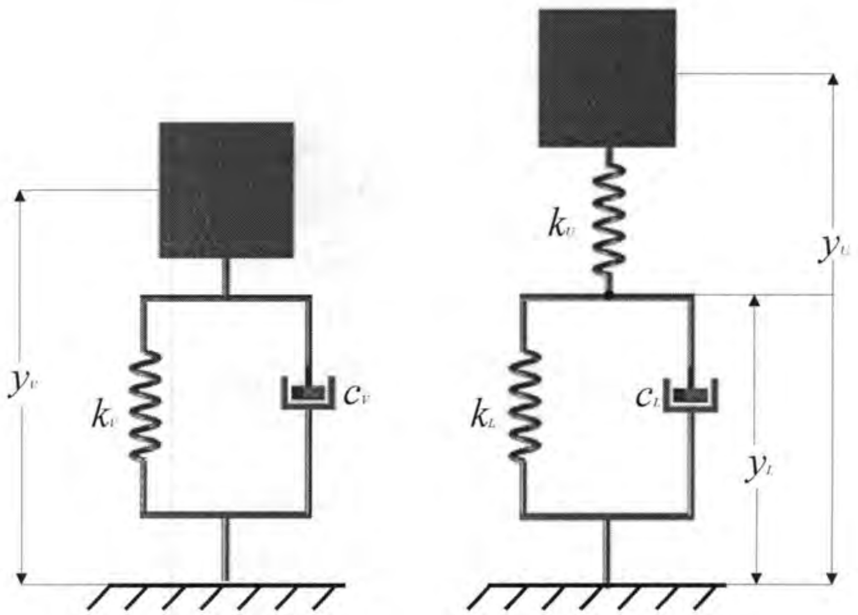
Ground Reaction Forces. Subjects landed on a 0.60 x 0.40 m force platform (Kistler, Model 9281B, Amherst, NY 14228) with both feet. Ground reaction forces were collected for 95 ms at 1000 Hz. Prior to each landing, an A/D board in a computer triggered collection of force data when a beam of light entering a photo resistor located 10 cm above the force plate was disrupted by each subject's feet. This provided data where the measured force should be zero. The force data were not filtered. If the pre-contact data were nonzero, then these pre-contact values were subtracted from every force value in the respective trial to correct for a small amount of drift in the force plate transducers. A trial was "acceptable" when both feet were completely on the force platform from initial contact to standing at rest. Trials were excluded if the subject made contact with any surface other than the force platform upon landing and if the computer was not triggered to record ground reaction force data before initial contact. Only 80 of the 1300 trials were excluded.

Mechanical Models. Center of mass displacement data and vertical ground reaction force data were fit with the Voigt (Figure 3.6a) and standard linear solid (Figure 3.6b) models. The Voigt model consists of a mass resting on top of a spring in parallel with a damper. The standard linear solid consists of a mass resting on a spring in series with the parallel spring and damper setup in the Voigt model. Center of mass acceleration measured from the force plate was integrated twice to determine center of mass displacement. For fitting purposes, displacement data were weighted from initial contact to 0.1 s past the point of maximal center of mass displacement. Since the subjects were not required to stand back up immediately after landing, significant variance occurred towards the end of the displacement landing trace. Thus, this part of the response was not included in the least squares fit. The force data were weighted from initial contact to 0.05 s past the maximum ground reaction force. The equations of motion used to fit the force and displacement data for the Voigt model and standard linear solid are shown in equation 3.1 and equations 3.2 and 3.3 respectively:

$$m\ddot{y}_V + c_V\dot{y}_V + k_V y_V = -W \tag{3.1}$$

$$m\ddot{y}_U + k_U\left(y_U - y_L\right) = -W \tag{3.2}$$

$$c_L\dot{y}_L = k_U y_U - \left(k_U + k_L\right)y_L \tag{3.3}$$



a. Voigt b. Standard Linear Solid

Figure 3.6. Dynamic models.

The equations of motion were integrated using the *ode45* solver in MATLAB v. 5.3 (Mathworks, Inc., Natick, MA 01760). The ordinary differential equation solver is based on an explicit Runge-Kutta formula that needs only the solution at the preceding time point, i.e. initial conditions, to perform the integration. Natural frequency (equation

3.4) and damping ratio (equation 3.5) were calculated using stiffness and damping predicted from Voigt model fits of the displacement data.

$$\omega_n = \sqrt{\frac{k_V}{m}} \tag{3.4}$$

$$\zeta = \frac{c_V}{2 \cdot \omega_n \cdot m} \tag{3.5}$$

Data Analysis. The data from several subjects had standard deviations in stiffness and damping that were larger than the mean for various variables suggesting non-normal data. The non-normality could be seen in scatter plots and often consisted of up to 14 extreme outliers defined using boxplots. However, there was no consistency in the distribution of the data between subjects and variables, therefore medians have been reported in addition to means and standard deviations. In addition, since the outliers were predicted using acceptable force and displacement data, no data were excluded except those where the computer froze in the fitting procedure. During a few of the Voigt model displacement fits the computer froze and stopped all computation preventing a fit from being performed. The reason for the freezing is still unclear. However, the trials that did freeze the computer all had similarities in the displacement trace at the last weighted point. If the last weighted point fell in a region of the trace that was linear following maximum displacement the fitting procedure tended to freeze. However, if the weighted region was moved closer to or further away from this linear region, where the trace was curved slightly, then the software carried out the fit. The fitting routine is based on error magnitude and it is thought that at this linear area the software was having difficulty fitting to a straight line region coming from a sinusoidal trace. Instead of manually changing the weighted region for those trials that caused the computer to freeze, the trials were excluded in order to maintain consistency in the fitting procedure.

The goodness of fit of each model to the force and displacement data was determined for each trial by calculating the standard deviation of the error (SDE) between the experimental data and predicted data along the weighted region of the trace. Analysis of variance was used compare the goodness of fit between models in both fit types.

3.4 Results

The standard deviation of the error (SDE) provides an indication of how well the two models represent the experimental data. The SDE in the typical displacement fit (Figure 3.7) using the Voigt model was 0.003 ± 0.001 m (Table 3.2). The SDE in the typical displacement fit using the standard linear solid model (Figure 3.8) was 0.001 ± 0.001 m (Table 3.3). Statistically, the standard linear solid fit the displacement traces better than the Voigt model ($p<0.001$). However, on a practical level, the standard deviation of the error in the Voigt model was only 3.0 mm. Qualitatively there appeared to be no difference between how well the two models fit the displacement trace. When performing Voigt model fits of displacement data the computer froze on several trials causing fewer trials ($n = 946$) to be fit compared to the standard linear solid ($n = 1131$).

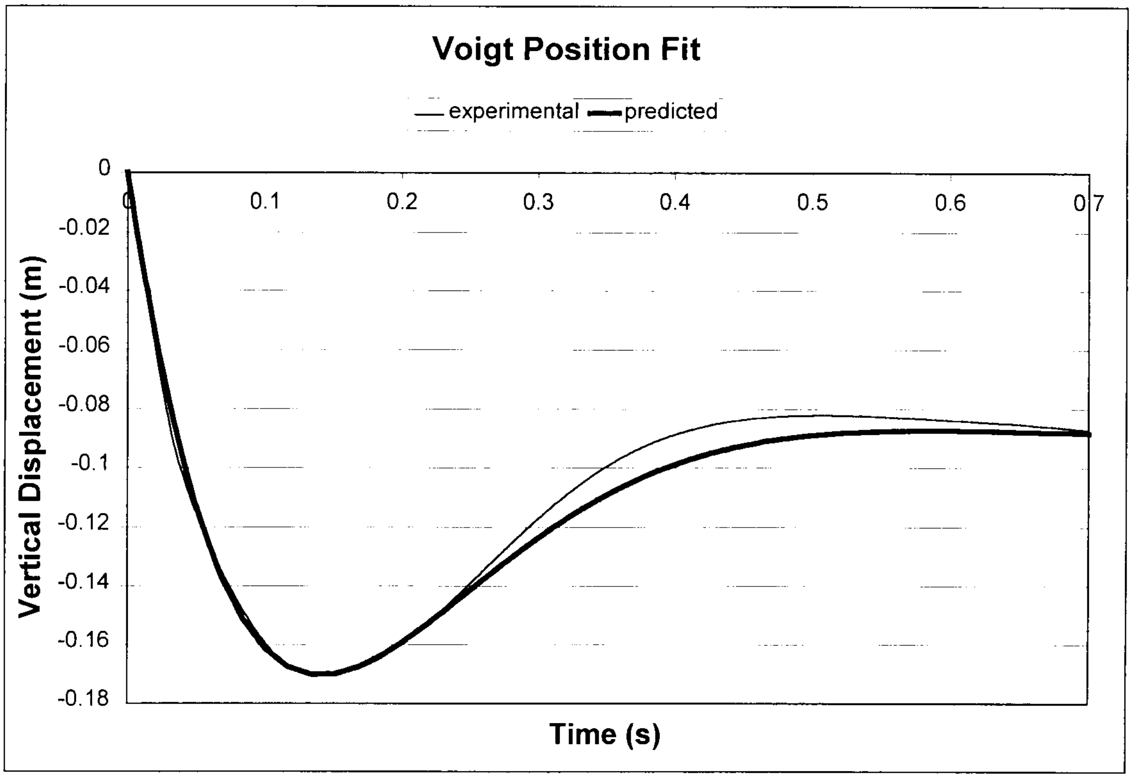


Figure 3.7. Typical Voigt displacement fit.

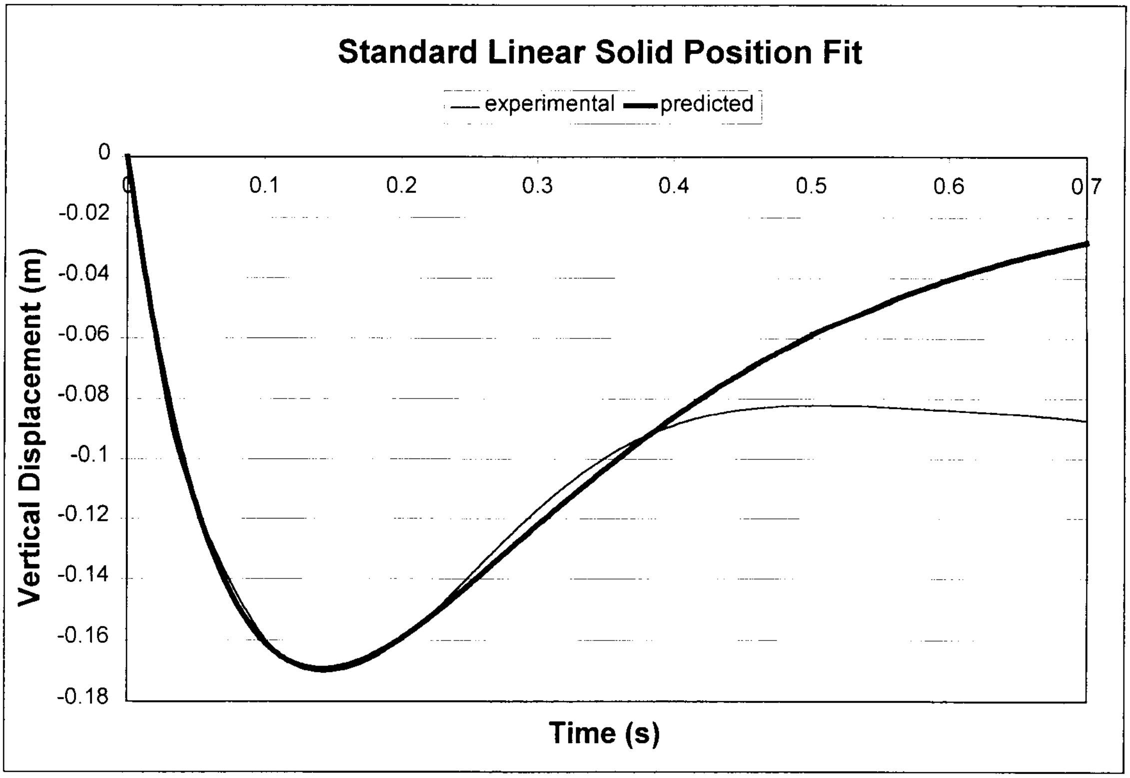


Figure 3.8. Typical standard linear solid displacement fit.

Table 3.2. Voigt displacement fit summary data.

Subject	cv ± SD	kv ± SD	Wn ± SD	Z ± SD	SD Error ± SD	n
Subj 1	436.3 ± 90.3	3302.6 ± 1523.9	10.5 ± 2.0	0.72 ± 0.08	0.002 ± 0.000	96
median	425.7	2998.4	10.2	0.71	0.002	
Subj 2	467.2 ± 265.2	4876.8 ± 3799.6	12.9 ± 3.3	0.64 ± 0.11	0.002 ± 0.001	100
median	425.0	4280.0	12.5	0.64	0.002	
Subj 3	546.5 ± 565.2	3805.9 ± 5962.1	9.7 ± 7.1	1.40 ± 1.70	0.003 ± 0.002	61
median	365.6	2053.4	8.8	1.08	0.002	
Subj 4	896.3 ± 696.6	9790.2 ± 9342.4	16.2 ± 8.2	1.07 ± 1.21	0.003 ± 0.002	49
median	593.3	5295.4	13.3	0.76	0.002	
Subj 5	503.5 ± 473.9	4125.5 ± 7591.9	9.7 ± 7.8	3.12 ± 8.11	0.003 ± 0.002	44
median	350.4	1953.6	8.6	0.81	0.002	
Subj 6	487.9 ± 102.6	4564.1 ± 1681.4	11.1 ± 2.0	0.61 ± 0.07	0.003 ± 0.000	99
median	461.1	4241.6	10.9	0.61	0.003	
Subj 7	623.3 ± 377.6	5219.1 ± 5691.4	11.8 ± 5.5	0.89 ± 0.41	0.003 ± 0.001	50
median	502.8	3282.1	10.3	0.81	0.002	
Subj 8	990.6 ± 933.1	9572.8 ± 10943.0	14.4 ± 8.9	0.91 ± 0.33	0.004 ± 0.002	41
median	506.3	2922.5	9.3	0.74	0.003	
Subj 9	871.7 ± 473.5	10720.1 ± 6495.8	15.0 ± 4.1	0.63 ± 0.14	0.002 ± 0.001	93
median	710.1	8419.3	13.8	0.60	0.002	
Subj 10	611.4 ± 377.9	5308.2 ± 5196.9	9.9 ± 3.7	0.62 ± 0.10	0.003 ± 0.001	95
median	489.3	3684.7	8.8	0.60	0.002	
Subj 11	615.5 ± 419.3	6620.0 ± 5391.9	13.1 ± 4.4	0.64 ± 0.14	0.003 ± 0.001	93
median	481.5	4801.1	11.7	0.60	0.003	
Subj 12	1084.9 ± 505.4	10548.4 ± 8044.1	13.1 ± 4.9	0.80 ± 0.37	0.003 ± 0.001	82
median	919.4	7837.9	12.1	0.75	0.003	
Subj 13	1038.1 ± 830.6	10157.0 ± 10919.9	13.5 ± 8.6	1.05 ± 0.56	0.003 ± 0.001	43
median	584.7	5102.8	11.3	0.87	0.003	
Total	674.5 ± 512.9	6577.5 ± 6781.8	12.3 ± 5.5	0.89 ± 1.89	0.003 ± 0.001	946
median	482.7	4368.9	11.3	0.68	0.002	

The SDE in the typical force fit using the Voigt model (Figure 3.9) was 580 ± 269 N (Table 3.4). The SDE in the typical force fit using the standard linear solid model (Figure 3.10) was 493 ± 245 N (Table 3.5). The initial peak in the standard linear solid force trace was fit through the two initial force peaks on the ground reaction force trace resulting in predicted maximum force values 13% less than the actual peak force. The under prediction equates to approximately 0.7 times body weight and is most likely a result of trying to model the initial two force peaks in the experimental data with a single degree of freedom model.

Specific stiffness and damping values are much different in fitting the impact portion of the force data compared to fitting displacement data. This reinforces the apparent difference between the response of the body at initial contact versus the global response to the drop landing event. In the displacement fits, the median value of the stiffness of the spring in parallel with the dashpot for the Voigt (k_V) and standard linear solid (k_L) was 4368 N/m (mean ± SD) and 1206 N/m respectively. Damping for the Voigt (c_V) and standard linear solid (c_L) was 483 N·s/m and 404 N·s/m respectively. Stiffness of the series spring in the standard linear solid (k_U) was 48.6 kN/m. The median natural frequency and damping ratio of the children in this study was 11.3 rad/s and 68% respectively (Table 3.2).

In fitting the forces, the median value of the stiffness of the spring in parallel with the dashpot for the Voigt (k_V) and standard linear solid (k_L) was 21402 N/m (mean ± SD) and 3.0 N/m respectively. Damping for the Voigt (c_V) and standard linear solid (c_L) was 516 N-s/m and 1384 N-s/m respectively. Stiffness of the series spring in the standard linear solid (k_U) was 69.5 kN/m.

Table 3.3. Standard linear solid displacement fit summary data.

Subject	cl ± SD	kl ± SD	ku ± SD	SD Error ± SD	n
Subj 1	389.4 ± 39.3	1436.4 ± 430.3	123472.3 ± 126920.9	0.001 ± 0.0002	87
median	391.7	1373.5	74280.3	0.001	
Subj 2	381.5 ± 35.5	1295.5 ± 382.5	38721.9 ± 11519.8	0.001 ± 0.0002	100
median	382.5	1284.6	35553.6	0.001	
Subj 3	236.9 ± 61.4	479.7 ± 558.5	62168.0 ± 60119.8	0.001 ± 0.0013	82
median	227.3	249.6	39925.9	0.001	
Subj 4	329.4 ± 62.5	595.6 ± 423.3	121775.7 ± 240734.6	0.001 ± 0.0012	79
median	326.6	498.2	64514.6	0.001	
Subj 5	292.9 ± 49.4	738.8 ± 756.2	381326.7 ± 1366771.0	0.001 ± 0.0002	67
median	292.9	514.5	89770.9	0.001	
Subj 6	456.4 ± 55.1	1837.2 ± 586.9	52403.5 ± 20930.2	0.001 ± 0.0002	98
median	452.4	1798.0	48167.3	0.001	
Subj 7	392.7 ± 61.2	761.9 ± 452.8	132003.7 ± 175935.6	0.001 ± 0.0012	53
median	392.0	730.6	62440.1	0.001	
Subj 8	287.5 ± 46.0	567.7 ± 287.0	64540.2 ± 96341.2	0.001 ± 0.0003	98
median	279.4	525.5	41686.5	0.001	
Subj 9	532.6 ± 61.1	1451.2 ± 568.5	37550.2 ± 8819.5	0.001 ± 0.0002	98
median	536.8	1417.2	36675.7	0.001	
Subj 10	506.1 ± 62.9	2267.8 ± 662.5	98177.9 ± 97038.9	0.001 ± 0.0002	96
median	498.2	2333.7	74792.4	0.001	
Subj 11	391.3 ± 38.0	1258.8 ± 431.5	44727.6 ± 60735.8	0.001 ± 0.0002	99
median	388.2	1202.6	33676.7	0.001	
Subj 12	763.0 ± 228.6	1791.3 ± 1350.3	147432.9 ± 189459.0	0.002 ± 0.0034	93
median	740.8	1627.7	87296.2	0.003	
Subj 13	410.6 ± 81.5	582.6 ± 475.4	78664.4 ± 70164.2	0.002 ± 0.0004	81
median	403.0	453.1	56153.1	0.002	
Total	429.4 ± 160.3	1267.6 ± 844.5	81103.4 ± 132508.0	0.001 ± 0.0012	1131
median	403.8	1206.7	48616.9	0.001	

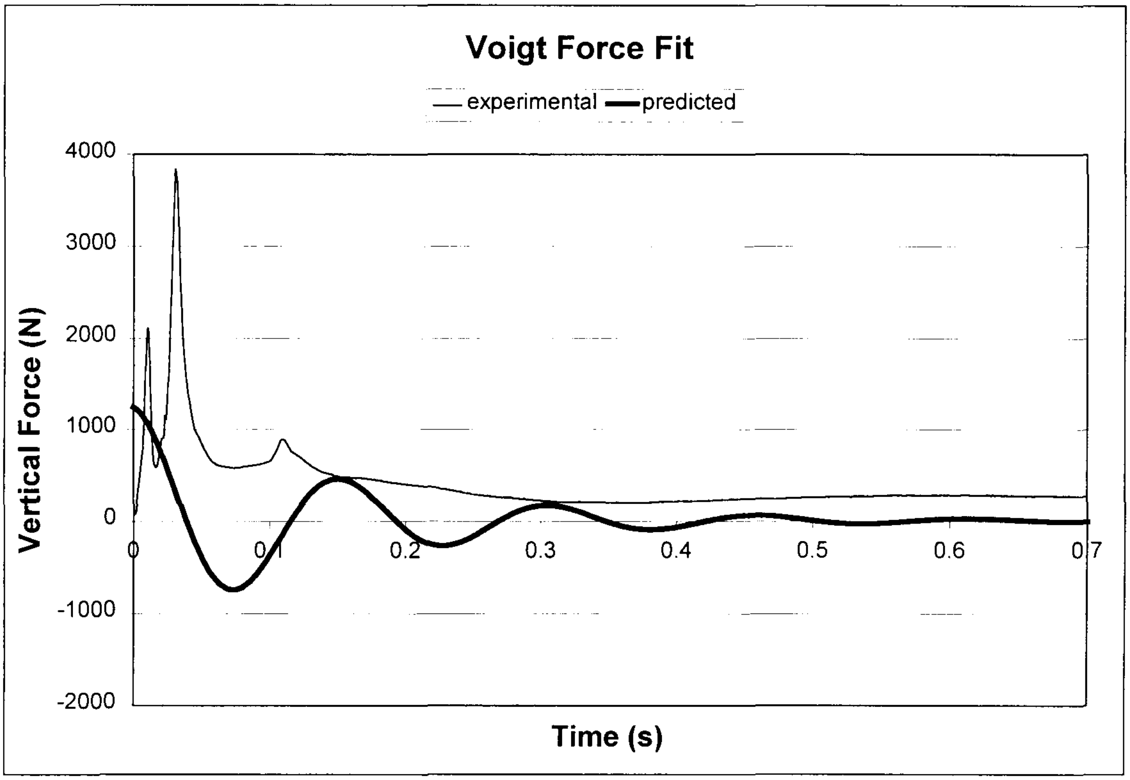


Figure 3.9. Typical Voigt force fit.

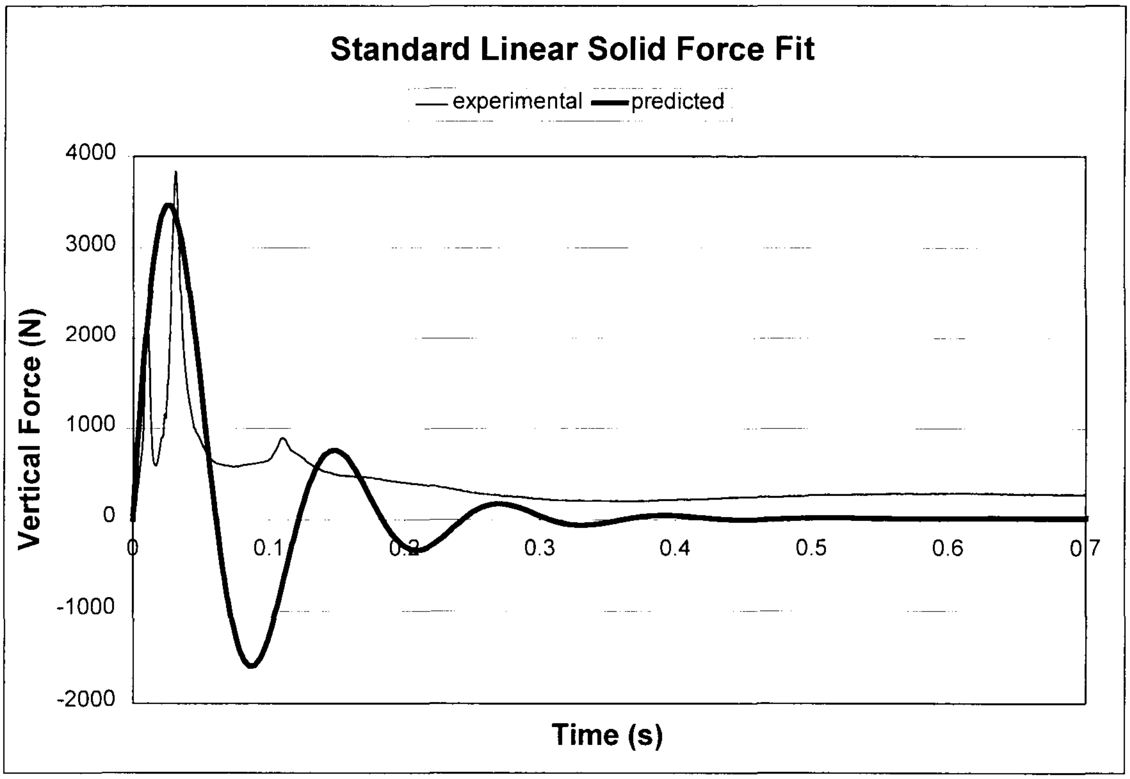


Figure 3.10. Typical standard linear solid force fit.

Table 3.4. Voigt force fit summary data.

Subject	cv ± SD	kv ± SD	SD Error ± SD	n
Subj 1	448.4 ± 58.7	40825.7 ± 13507.5	634.7 ± 121.7	99
median	443.1	40179.5	643.1	
Subj 2	433.2 ± 51.5	17528.4 ± 6020.2	412.5 ± 84.7	100
median	427.5	16697.1	418.4	
Subj 3	412.2 ± 75.0	9532.6 ± 7988.2	296.9 ± 62.5	92
median	395.5	6286.2	281.4	
Subj 4	469.2 ± 89.8	16018.5 ± 7985.7	391.9 ± 100.7	94
median	468.9	14691.0	376.5	
Subj 5	450.3 ± 164.4	24785.7 ± 12685.3	424.7 ± 102.1	69
median	425.9	21130.8	399.3	
Subj 6	506.6 ± 51.6	23864.9 ± 6748.8	569.3 ± 102.7	100
median	508.8	23623.2	579.8	
Subj 7	534.6 ± 118.9	31027.2 ± 12567.5	591.0 ± 139.9	68
median	511.2	29683.3	586.6	
Subj 8	403.5 ± 51.3	7950.1 ± 4229.5	295.4 ± 72.7	99
median	394.2	6809.7	271.0	
Subj 9	694.8 ± 100.9	22548.7 ± 9140.7	652.4 ± 160.5	98
median	686.2	20894.8	649.6	
Subj 10	788.2 ± 64.3	58299.0 ± 20356.3	976.6 ± 154.9	99
median	779.8	55561.7	967.0	
Subj 11	526.1 ± 50.0	21047.1 ± 5518.0	496.5 ± 84.4	100
median	523.5	20794.7	493.3	
Subj 12	879.2 ± 151.3	45200.4 ± 18050.7	1034.6 ± 243.0	100
median	836.6	40712.2	985.1	
Subj 13	629.3 ± 81.3	23623.3 ± 11538.3	589.1 ± 140.9	90
median	621.2	19976.8	565.9	
Total	563.6 ± 175.3	26228.3 ± 18745.5	580.0 ± 269.1	1208
median	515.6	21402.1	529.4	

Table 3.5. Standard linear solid force fit summary data.

Subject	cl ± SD	kl ± SD	ku ± SD	SD Error ± SD	n
Subj 1	2320.6 ± 1124.6	4301.6 ± 6650.5	77785.4 ± 17590.4	710.8 ± 175.1	99
median	2121.6	11.1	76242.3	719.6	
Subj 2	859.9 ± 484.5	8625.1 ± 7641.7	102422.9 ± 140666.1	369.0 ± 98.5	100
median	663.4	8743.1	52583.3	356.0	
Subj 3	618.7 ± 321.3	5192.6 ± 5876.2	111178.5 ± 83842.6	264.9 ± 99.1	92
median	504.7	3309.3	63628.7	252.4	
Subj 4	1656.1 ± 6843.6	5566.6 ± 6828.1	104697.6 ± 72222.0	446.2 ± 121.4	95
median	868.6	1554.0	71208.0	449.1	
Subj 5	1352.1 ± 837.1	3329.4 ± 7869.6	91301.7 ± 65630.9	473.0 ± 129.7	69
median	1175.5	0.0	68703.5	464.1	
Subj 6	1699.0 ± 604.7	2863.7 ± 6565.5	89748.9 ± 159540.3	377.8 ± 101.3	100
median	1827.9	0.0	53953.1	363.9	
Subj 7	1893.1 ± 756.2	244.6 ± 1490.3	65622.6 ± 16018.8	457.9 ± 159.6	68
median	1762.8	0.0	63327.7	461.1	
Subj 8	617.9 ± 346.8	5005.1 ± 4434.6	153422.2 ± 251223.5	244.0 ± 94.3	99
median	471.2	5552.9	58422.7	225.0	
Subj 9	1027.8 ± 557.9	15365.1 ± 9032.0	216972.6 ± 230043.8	583.4 ± 166.5	98
median	754.7	17544.7	169148.2	575.5	
Subj 10	3418.4 ± 1200.9	4099.6 ± 7075.4	99425.1 ± 19801.8	940.9 ± 225.3	99
median	3171.1	32.8	97316.8	919.3	
Subj 11	1625.5 ± 341.9	67.4 ± 368.5	47968.8 ± 7702.0	304.2 ± 76.5	100
median	1577.4	0.0	48728.6	294.7	
Subj 12	1504.3 ± 1052.7	1333.0 ± 1534.6	134841.6 ± 181144.5	252.2 ± 340.5	100
median	2828.0	0.0	93880.6	658.4	
Subj 13	1572.6 ± 651.9	2798.5 ± 6069.9	73851.6 ± 61830.8	525.2 ± 144.8	87
median	1426.2	0.0	60657.5	503.6	
Total	1666.7 ± 2193.7	4605.4 ± 7327.9	104667.6 ± 131183.5	493.4 ± 244.6	1206
median	1384.2	3.0	69554.4	450.3	

3.5 Discussion

In this study we performed experiments to answer two questions. First, can single degree of freedom models be used to characterize the stiffness and damping properties of children performing drop landings? By fitting the experimental data with data predicted from single degree of freedom dynamic models we were able to replicate center of mass displacement through maximum displacement, or the global response to the drop landing event, quite accurately (SLS SDE = 0.001 m, Voigt SDE = 0.003 m). However, while the standard linear solid out performed the Voigt model in fitting ground reaction force data (SLS SDE = 493 N, Voigt SDE = 580 N), the standard linear solid was unable to match both initial force peaks present in the experimental data. Instead the model tried to fit one predicted peak force through both initial force peaks in the experimental data (Figure 3.10). While this prevents us from predicting the complete time history of the force trace, the model still predicted the peak force to within 13%.

In the second question we ask, what are the stiffness and damping characteristics of children performing drop landings? As might be apparent from looking at the traces of center of mass displacement and ground reaction forces, fitting to the force required a stiffer system than fitting to the displacement trace. Comparing how the two models fit displacement data, the series spring (k_U) in the standard linear solid model was so stiff that it was practically a rigid link, making the standard linear solid act very similar to the Voigt model. However, when fitting force data, the standard linear solid model often had stiffness values of 0.0 N/m on the lower spring (k_L), vastly changing the configuration of the system compared to the Voigt model. The presence of the series spring (k_U) with the standard linear solid allowed the predicted force data to begin at 0.0 N. This is important because the children made contact with the force plate at an average velocity of 3.1 m/s. Given an initial velocity, the parallel combination of the spring and damper in the Voigt model could only predict nonzero initial forces (Figure 3.9)

Using simple single degree of freedom dynamic models to describe both the center of mass displacement and the ground reaction force characteristics of a drop landing event are unique to this study. The inclusion of the initial impact force data in stiffness calculations, which are normally omitted from research investigating stiffness of the leg or whole body, allowed the most likely characteristic for osteogenesis of drop landing to be simulated. The two distinct fit types provide two aspects of a child's response to landing. In addition, the number of trials (n=1220) available for use in testing goodness of fit greatly added to the credibility of the results.

There are limitations any time mechanical systems are used to model human characteristics. In using the mechanical models to describe human movement, it was assumed that the models represented the global system of muscles, ligaments, tendons, and other tissues in the human body. Also, while the single degree of freedom models in this study were chosen for simplicity, the calculations are still more complicated than those used for a linear mass spring model, making them more difficult to implement. A limitation of the standard linear solid is that, while being superior at fitting the peak forces compared to the Voigt model, the standard linear solid cannot emulate both initial force peaks. Instead the standard linear solid created one peak through the two initial force peaks (Figure 3.10), which underestimates the actual stiffness of the system during initial impact. Lastly, the weighting methods used to fit the models to the data were chosen based on the rationale that the end of each trial was not important to characterize mechanically.

The children in this study had a median damping ratio of 68% and natural frequency of 11.3 rad/s. As the natural frequency of the system increased from the median value, or when the children landed more stiffly, the damping ratio appears to increase linearly (Figure 3.11). As the natural frequency decreased from the median, or when the children landed more softly, the damping ratio appears to increase exponentially. Using a more sophisticated two degree of freedom model, similar to combining two Voigt systems in series, Mizrahi and Susak (1982), using the lower mass,

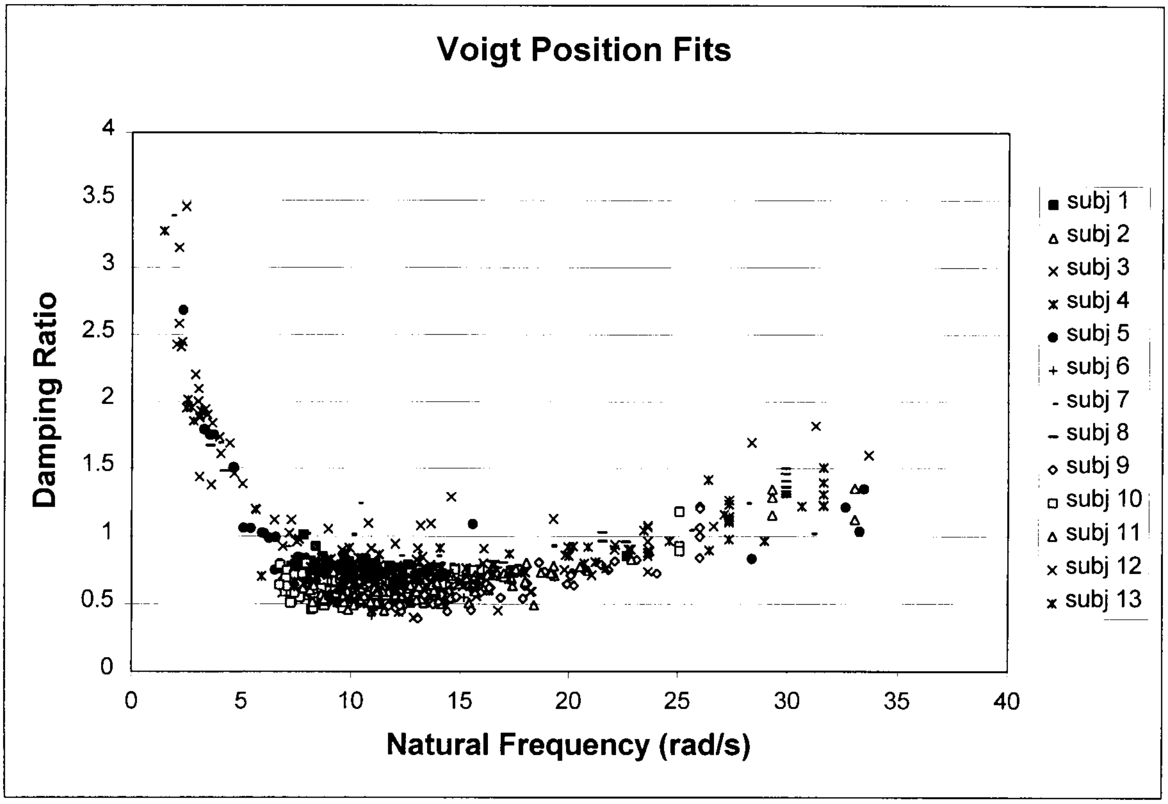


Figure 3.11. Damping ratio v. natural frequency from Voigt model displacement fits.

spring and damper, calculated damping ratios of 52% and 67% and natural frequencies of 10.8 and 9.1 rad/s for two subjects respectively when modeling impact forces from a landing. Using the same two degree of freedom model, Ozguven and Berme (1988) assumed a damping ratio of 50% for the upper mass, spring and damper when developing a dynamic system to predict impact forces during landing from a jump. They reported the average natural frequency for the lower mass, spring and damper to be 13.1 rad/s. Both research groups used adult subjects, whereas our research involved prepubertal children. The calculated natural frequency of the children falls in between the values reported by Mizrahi and Susak (1982) and Ozguven and Berme (1988). The calculated damping ratio of the children is slightly higher than reported in adults. It is important to note that a direct comparison between the Voigt model in our study to separate modes of the two degree of freedom model used by Mizrahi and Susak (1982) is not completely appropriate since characteristics such as stiffness, damping, natural frequency and damping ratio in one mode does not represent the characteristics of the entire system. However, the comparison was made as those were the only studies found to report damping ratio and natural frequency during landing activities. As of this writing, few comparisons among the stiffness and damping values reported here can be made to other studies.

Simple linear mass spring models are useful in describing stiffness characteristics of the human body during activities where the mass displaces in a sinusoidal fashion. However, in using the linear mass spring model researchers have generally assumed that maximum center of mass displacement coincides with maximum ground reaction forces, an assumption that is clearly not true in drop landings (Figure 3.3). Dyhre-Poulsen, et al. (1991) attempted to model the sharp force peaks at impact with center of mass displacement by calculating an instantaneous stiffness. Using force data collected from subjects performing drop landings from a height of 60 cm, stiffness was calculated by first order finite differences. This approach involves use of a linear mass spring model to represent jumpers landing from a height. While their paper was one of the few to make use of a single degree of freedom model to fit the force peaks at initial impact, calculating stiffness for each increment in time from a drop landing produces many different stiffness values throughout one landing, making it difficult to develop a practical interpretation of the results. In addition, the center of mass displacement trace observed during drop landings appears to be damped, implying that a damper should be added to the mass spring model to represent drop landings most appropriately. Such a model has been used previously to represent drop landings and the response of human tissue to impact (Minetti et al., 1998; Robinovitch, et al., 1997).

Using the Voigt model adds damping to the linear mass spring model, which attempts to account for inelastic components of the human body. Using a mechanical model to fit vertical displacement data allows for calculations of stiffness and damping that describe how the body responds globally to drop landings. When modeling peak forces, the Voigt model has been shown to predict forces accurately at impact velocities between 1.2-1.9 m/s (Robinovitch, et al., 1997). However, at impact velocities greater than 2.2 m/s the standard linear solid better predicted impact forces, in keeping with our results. Splitting the landing into an impact specific signal (force data) and a global response signal (displacement data) allows for a more clear picture of what is actually happening and is of particular interest in bone research. Specifically, modeling the force data with a mechanical model provides an estimate of mechanical characteristics of the body during a reflexive muscular response to initial impact, whereas modeling displacement data provides a more general response of the body during the entire landing event.

Using mechanical models to represent human motion allows researchers to make predictions about what type of motion and forces might occur given specific initial conditions. The peak forces are important to model because they tend to represent reflexive muscle stiffness of the lower extremity, which is the portion of the signal most likely associated with osteogenesis because of the high magnitude forces and fast loading rate. In addition, this can be done without direct access to a force measuring device. If one model can be found that can accurately and simply represent the human body during several activities such as walking, running and drop landing then researchers desiring a specific ground reaction force and/or displacement profile could use the model to determine whether or not the activity might actually provide the stimulus that is needed for osteogenesis before implementing the activity in an exercise program.

3.6 Symbols

W	Vertical ground reaction force.
k_V	Stiffness of spring in Voigt model.
k_U	Stiffness of series spring in standard linear solid model.
k_L	Stiffness of lower spring in standard linear solid model.
c_V	Damping in Voigt model.
c_L	Damping in standard linear solid.
y_V	Position of mass in Voigt model.
$\dot{y}_V$	Velocity of mass in Voigt model.
$\ddot{y}_V$	Acceleration of mass in Voigt model.
y_U	Position of mass in standard linear solid model.
$\ddot{y}_U$	Acceleration of mass in standard linear solid model.
y_L	Position of connection between series spring and lower spring/damper system in standard linear solid model.
$\dot{y}_L$	Velocity of connection between series spring and lower spring/damper system in standard linear solid model.
ω_n	Natural frequency.
ζ	Damping ratio.

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CHAPTER 4
CONCLUSION AND RECOMMENDATIONS FOR FUTURE RESEARCH

All results from studies designed to investigate the effects of particular exercises on bone mass in humans are significant. While the statistics may not always show significant changes in bone mass caused by participation in the respective exercise intervention, the field of bone research is one where non-significant results are very helpful. However, a problem arises when research investigating the relationship between exercise and bone mass uses a program comprised of multiple activities that have not been properly evaluated in the potential for osteogenic effects. Specifically, Korht et al. (1997) looked to investigate the differences between two exercise regimens on osteogenesis. One exercise program was based on ground reaction forces, which included activities such as walking, running and stair climbing. The second exercise regimen was based on joint reaction forces resultant from weight lifting and rowing. One inherent problem is that if there is a change in bone mass at the hip or the spine it is impossible to tell which individual component of either exercise regimen was responsible for the change. In addition, while the authors rationalize the intervention through animal studies that emphasize magnitude of loading, rate of loading and frequency of loading as key considerations, no attempt was made to estimate what the load magnitudes and loading rates were in their study. Not knowing the contributions of individual exercises to osteogenesis makes it difficult to advance our knowledge of exercise prescription for osteoporosis prevention.

For the time being research examining the relationship between exercise and bone mass should stick to one carefully controlled activity so that the potential individual contribution of that activity to osteogenesis can be determined. Bone responds quite differently to different types of stimuli. Carefully controlled research in bone biology has demonstrated that greater magnitudes of applied forces are more osteogenic than lower magnitude forces (Rubin and Lanyon, 1987). Researchers have also reported that faster rates of bone deformation are more osteogenic than slower rates of deformation. In addition, the frequency of loading does not seem to play a key role. Rubin and Lanyon (1987) reported that 4 loading cycles a day were enough to maintain bone mass in a turkey ulna whereas 36 loading cycles increased bone mass and did not have significantly different effects on osteogenesis compared to 1800 loading cycles per day. Knowing the information reported from carefully controlled animal studies, researchers should be able to conclude with confidence that activities with large magnitude force characteristics and large strain rates applied at moderate frequencies should be osteogenic in humans. Therefore, the force characteristics of the activities used in bone research exercise intervention studies should be known before the intervention is ever implemented. In addition, all bone research exercise intervention studies should descriptively report the characteristics of the activities or cite research that has quantified the activities. If the bone research field, and particularly the exercise related bone research field is ever to solve the puzzle of preventing osteoporosis through exercise, researchers must first thoroughly define the pieces of the puzzle. Knowing the characteristics of individual activities and whether or not the activities had any influence on bone mass allows future researchers and osteoporosis prevention advocates to narrow down the types of activities that can be implemented in exercise programs worldwide.

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APPENDICES

APPENDIX A
EFFECTS OF DETRAINING ON GROWING BONES

1. Brief Project Description

Researchers suggest that adding bone during childhood is the best preventive measure for the future protection against fractures, and osteoporosis. It has been estimated that a 3% increase in bone mineral density can reduce fracture risk by 20%. Our laboratory developed and implemented a unique exercise program designed specifically to increase bone mass at the hip in prepubescent children. We hypothesized that we would observe significant increases in bone mass at the hip in those children who performed jumping activities compared to those who performed low impact flexibility exercises. A total of 34 children participated in this program during the 1997-98 school year at Harding Elementary School. Results from this pilot study revealed that children who performed 100 jumps off 24-inch boxes three times per week had a *5.6* % greater increase in bone mineral content at the hip than did controls. Findings from this pilot study provide preliminary evidence that in fact impact activities are important for building bone.

In the current proposed study we will examine the effects of detraining (removal of jumping) on growing bone, and quantify the range of forces associated with the specific jumping exercises. First, we will measure bone mineral density in those children who completed the seven-month exercise intervention during the 1997-98 school year, ten months after the completion of the exercise program. This will provide valuable information on whether or not the increases in bone mass found in the jumping group were maintained. Furthermore, a complete assessment of the range of forces over 100 jumps will be examined in order to quantify the stimulus that was associated with increases in hip bone mass. It is expected that forces will be from 5-8 times body weight and decline slightly with fatigue.

2. Methods and Time Line

Testing sessions will take place over a period of three weeks starting the second week of April 1999. Each participant will only be required to come in once to the testing site. Testing will take place at Oregon State University at the Bone Research Laboratory and the Biomechanics Laboratory, both located in the Women's Building. Bone mineral density will be measured in the Bone Research Laboratory, whereas leg stiffness and ground reaction forces will be measured in the Biomechanics Laboratory. Testing will begin in the Bone Research Laboratory and conclude in the Biomechanics Laboratory. Details of the testing are described below.

A. Bone Research Laboratory Testing Measurements:

1. All participants will undergo the same testing measurements performed in the study entitled "The Effects of Jumping on Growing Bone." These tests include a bone mineral density scan, skinfold testing, completion of a physical activity and nutrition questionnaire. These tests are described in the informed consent form.

B. Biomechanics Laboratory Testing Measurements:

1. After a 10 minute warm-up of light aerobic activity and dynamic stretching each participant will perform 100-drop landings from a height of twenty-four inches in a period of approximately 10 minutes. Each landing should be as stiff as possible with hands on hips. This landing style is the same landing style used in the study entitled "The Effects of Jumping on Growing Bones" completed last year.

2. Estimation of leg stiffness: In order to assess changes in leg stiffness ground reaction forces from each drop landing must be recorded. Upon landing, each participant will contact a force platform (Kistler, 9281B) with their left foot and a wood floor adjacent to the force platform with their right foot. The gap separating the force plate and the floor is approximately 4mm. A thin (< lmm) surface will cover the force plate and adjacent wood floor to prevent any biasing that may occur with the knowledge of landing on a force platform. A white line will be placed on the landing surface marking the separation between the force plate and wood floor. The children will be asked to land with their feet on either side of the white line.

3. Estimation of Joint Reaction Forces: Joint reaction forces will be estimated using inverse dynamics. Inverse dynamics calculations require ground reaction force estimates and the positions of body landmarks through a complete cycle of the exercise task described above. The force plate will be used to collect ground reaction force estimates. A high-speed video camera will be used to collect the body landmark positions. One-inch diameter reflective markers made from 3M retro-reflective tape will be placed on the left side of the body at the following landmarks: fifth metatarsal, heel, toe, knee, femoral condyle, and shoulder. Each participant will wear black Lycra shorts and tank tops to minimize marker movement caused by the movement of loose clothing. Knee and shoulder markers will be placed directly on the skin. The reflective tape has the same adhesive qualities of a Band-Aid strip and will not cause any discomfort when removed.

3. Benefits and Risks from Participation

A. Benefits

Each child will receive valuable information regarding his/her bone mineral density, body composition, and muscular power as a result of participating in this study. The assessment of bone mineral density typically cost $100 for each bone scan, and this is free of charge.

B. Risks

Participants will be exposed to a very low dose of radiation from the bone densitometer used to measure bone mineral density. The maximum radiation received from a regional hip or spine scan is 1-4 μSv, equivalent to about 1/10th of a standard chest x-ray. Thus, the cumulative dose is slightly less than the amount of radiation an average individual receives in one day from background sources such as the sun.

No injuries were incurred during the pilot study in those children who performed both the jumping and stretching exercises. The testing session may produce acute and delayed onset muscle soreness (24-48 hours after exercise). Muscle soreness will be minimized by the inclusion of a 5-10 minute warm-up and cool-down. There is also a slight chance of injury due to accident. To minimize accidental injury the landing area will be free of any obstacles within a 2-meter radius. All jumping exercises performed in the Biomechanics Laboratory on the day of testing will be closely monitored by trained personnel.

4. Participant Population

Participants for this study will include apparently healthy prepubescent girls and boys between the ages of 7 and 10 years. Participants will include 34 boys and girls who completed the 7-month jumping exercise program between September 1997 and June 1998 at Harding Elementary School.

5. Informed Consent

Refer to attached informed consent.

6. Method of Obtaining Informed Consent

Parents of each potential participant will be contacted by telephone, and given a verbal description of the study. Interested persons will be scheduled for an appointment at which time the informed consent form will be reviewed. Both the parent and child will be required to sign the informed consent form before performing any testing measurements. All participants will be provided with a copy of the informed consent to keep.

7. Confidentiality

Participants will be assigned a code number, which will be used on all questionnaires and computer output, and will be stored in a separate file. Only the investigators will have knowledge of each participant's name and code number.

EFFECTS OF DETRAINING ON GROWING BONES

Informed Consent Form

INTRODUCTION and STUDY DETAILS

My child has been invited by Dr. Christine Snow (Principal Investigator) to return for testing in April (1999). The purpose of having my child come back in for testing is to evaluate the effects of detraining on bone development and to quantify the forces that are associated with the jumping exercises that were performed during the 1997-98 school year. If my child was in the jumping group my child will also be asked to jump onto a force plate so that forces at the hip can be measured. Explanations of the testing measurements that will be used are explained below.

MEASUREMENTS

As the parent I will be asked to bring my child in for testing in April (1999) to the Oregon State University Bone Research Laboratory. The approximate time that it will take to complete all tests will be one hour and include the following:

Bone Mineral Density Testing: Bone mineral density testing will require my child to lie quietly on an x-ray table for a total of six minutes for the hip and spine scans.

Body Composition Testing: My child will have his/her body composition measured using skinfold calipers. My child and I have been shown how the calipers work, and it has been explained to me that this procedure will not hurt my child. Measurements will only be taken on the arm and shoulder. This procedure has been used in other children of this age group and has been demonstrated as a safe and reliable way to measure body fat.

Physical Activity Questionnaire: I will help my child complete a questionnaire that will ask questions about the types of activities my son/daughter participates in on a regular basis. My child will also be asked questions regarding the amount of TV watched on a weekly basis, and the types of organized sports in which my son/daughter may be involved.

Food Questionnaire: I will be recording my child's food intake on a food questionnaire that will take approximately 20 minutes to complete. This questionnaire will require me to answer questions based on the types of foods my son/daughter consumes on an annual basis.

Biomechanics Laboratory: If my child was in the jumping group, he/she will be asked to jump off a two foot box, 100 times onto a force plate. The force plate will record how hard my child lands on the floor. A video camera will film how my child lands on the floor. My child will have reflective tape placed on the heel, toe, ankle, knee, hip, and shoulder. The reflective tape can be easily removed. This tape allows for precise measurement of joint position during landing.

BENEFITS & RISK OF INJURY

My child will receive valuable information regarding his/her bone mineral density following the jumping or stretching program. The radiation dose is considered safe to administer and has been used in many studies. The amount of radiation that my child will receive is less than that from natural background radiation during a plane trip across the country, or from a day outside in the sun.

Force measurement will allow the investigators to associate the amount of stimulus required to change bone. The testing session may produce acute and delayed onset muscle soreness (24-48 hours after exercise). There is also a slight chance of injury due to accident. I understand that the University does not provide a research subject with compensation or medical treatment in the event a participant is injured, or as a result of participation in the research project.

CONFIDENTIALITY

Confidentiality will be maintained for my child by a number coding system. Only the researchers will have knowledge of my child's name. I have been informed that the results of this study may be published in scientific literature, and that these data will not reveal the identity of my child.

INVESTIGATOR INFORMATION

I have been informed and understand that nature and purpose of this research study. The researchers have offered to answer any questions that I may have. I understand that my child's participation in this study is voluntary and that I may remove my child from the study at any time without sacrificing of benefits to which my child is entitled. Questions about the research or any aspect of my child's participation should be directed to Dr. Christine Snow at 737-6788, Robyn Fuchs at 737-5935, or Jeremy Bauer at 737-5933. Any further questions that I have should be directed to Mary Nunn, Sponsored Program Officer, OSU Research Office, 737-0670. I have read the above information and agree for my child to participate.

Subject Signature____ Date______

Parent/Guardian Signature__________________________________ Date______

Investigators Signature____________________________________ Date______

APPENDIX B
REVIEW OF LITERATURE

Why worry about bone?

Approximately 300,000 hip fractures attributed to osteoporosis are reported annually in men and women (National Osteoporosis Foundation, 1999). A hip fracture severely decreases an individual's ability to walk independently. Reduced mobility can lead to permanent disability, decreased quality of life, low self-esteem and eventually death. Total health care costs for treating hip fractures attributed to osteoporosis exceed $14 billion each year (Osteoporosis and Related Bone Diseases~National Resource Center, 1999). Of the prevention strategies, increasing bone mass during growth may be the most effective (Haapasalo et al., 1996).

Forces Associated with Exercise Induced Bone Growth

A dynamic load applied to bone is more osteogenic then a static load of the same magnitude or no applied load (Lanyon and Rubin, 1984; Rubin and Lanyon, 1987). In addition, the magnitude of loading appears to be more important than the frequency of loading (Whalen, Carter and Steele, 1988). Participation in dynamic high impact physical activities (i.e. those having ground reaction forces > 4 times body weight) is associated with greater bone mineral density at the hip (Fehling, Alekel, Clasey, Rector, and Stillman, 1995; Korht, Ehsani, and Birge, 1997; Taaffe, Robinson, Snow, and Marcus, 1997). Furthermore, researchers conducting cross sectional studies have reported that adults who participated in high impact activities during youth tend to have greater bone mineral density at the hip compared to adults who were less active during youth (Kirchner, Lewis, and O'Connor, 1996; Etherington, Harris, Nandra, Hart, Wolman, Doyle, Spector, 1996). Cross sectional reports provide evidence that accrual of bone mineral density during youth may improve peak bone mass, the maximum amount of bone gained in young adulthood. Greater peak bone mass will decrease the risk for hip fractures later in life. Greater ground reaction forces and muscle forces at the hip should translate into increased bone mass at the hip. In fact, bone mass in gymnasts, a population that is regularly exposed to high magnitude forces, is reported to be 35% above normal (Robinson, 1995).

Researchers at the Oregon State University Bone Research Laboratory have developed a highly specific exercise task designed to increase bone mineral density at the hip in prepubescent children. 34 boys and girls were assigned to either a jumping group (n=18) or a control group (n=16). The group of jumpers performed drop landings from a height of 61 cm, 100 times in one 10-minute session, three times a week with an emphasis on landing as stiff as possible. The control group performed stretches. After 7-months the jumpers exhibited a 5.6% greater increase in bone mineral content at the hip (Fuchs and Snow, 1998). Although a bone response was clear, the forces associated with the change in bone are not known.

Description of Landing Activities

Using principles of physics one can show that stiff landings, where joint movement is minimal, provide the greatest translation of ground reaction forces to the hip during drop landing exercises. Various landing strategies have been analyzed including those where the subjects had a knee angle range of motion greater than 90 degrees upon landing (soft) and with a knee angle range of motion less than 90 degrees upon landing (hard) (Devita, & Skelly, 1992). The knee angle is defined as the posterior angle between the thigh and shank. The ground reaction force impulse from the softer landings was 23% less than the stiff landing. The moments about the hip and knee joints were not different between soft and stiff landings. However the moment about the ankle was 25% greater in the stiff landing. Both landing styles, stiff and soft, required the fore foot to make contact with the ground before the rest of the foot. Thus, the landing style could be the reason greater moments at the hip were not observed in the stiff landings.

Ground reaction forces measured from gymnasts during landings have been measured at 11 times body weight from a height of 1.28 m, 8.8 - 14.4 times body weight when landing from a double back somersault and 8.2 - 11.6 times body weight when landing from dismounts on the horizontal bar (McNitt Gray, 1993; Panzer, Wood, Bates, and Mason, 1988; Ozguven and Berme, 1988). In 1986, Bobbert, Mackay, Schinkelshoek, Huijing, & van Ingen Schenau speculated that a hard, stiff landing will not be possible by the average person because of the high knee and hip extensor moments that must be exerted in order to counter the moments inherently created by the large forces at impact. A similar speculation was formed in 1993 when McNitt-Gray reported the kinetics of the lower extremity in both gymnasts and recreational athletes when landing from three different heights. The gymnasts tended to have greater peak moments at the hip extensors. Recreational athletes attenuated the reaction forces by using greater degrees of trunk flexion and knee flexion upon landing. The rationale for the difference was that a gymnast generally trains to land at high velocities from the high dismounts and is therefore equipped with the strength and experience to withstand high rotational forces at the joints. Recreational athletes that are untrained in landing rigidly appeared to fatigue more easily when performing multiple high impact landings due to the muscular exertion required by the hip and knee extensors and the ankle plantar flexors to maintain a rigid landing style. In 1997 Hoffman, Liebermann and Gusis reported that experienced skydivers landed with an average of 1 body weight more ground reaction force than novice skydivers even though there were no differences in maximal leg strength and power between groups. Therefore, there seems to be a trend of greater ground reaction forces in those who train (gymnasts, skydivers, etc.) to land compared to those who have not had training on landing techniques.

In 1990, Dufek and Bates reported differences in landing strategies among three subjects performing drop landings from three heights and three distances onto a force platform using three different landing techniques. Apart from the three different degrees of knee flexion required upon landing in the protocol, each subject varied in ground reaction force characteristics from the next in the same tasks suggesting large inter-individual variability. Schot, Bates and Dufek (1994) reported asymmetry in the magnitude of ground reaction forces between legs to be up to 14.8% suggesting that each leg is not subjected to exactly half of the measured ground reaction forces as is assumed in landing studies measuring forces from both legs with only one force plate. Asymmetry also becomes a problem in kinematic analyses using only one camera. Therefore, any analyses of landing should consider individual variability in landing style. In addition, researchers must be careful when describing how the ground reaction force is distributed to the body due to possible asymmetry during landing.

Kinematics and Kinetics of Children

An analysis of drop landing kinetics in children is central to understanding the dose response of bone to loading observed in the Oregon State University Bone Research Laboratory's jumping program. While some data in adults exist, measurement in children is necessary due to center of mass differences between adults and children. A child's center of mass is relatively higher in the body in relation to the location of an adult's center of mass (Jenson, 1989). A center of mass located higher in the body will potentially create greater moments about the hip during drop landing activities compared to a center of mass location lower in the body depending on skeletal alignment of the trunk over the hips upon landing. This greater torque over 100 trials might also have a magnified fatigue effect at the hip potentially causing the knee extensors and the ankle plantar flexors to accommodate the changes in the impact forces differently across trials. In addition, from direct observation of the children participating in the recent Oregon State University Bone Research Laboratory jumping program, it is clear that they become visibly tired after performing 100 sequential drop landings from 61 cm.

Schepens, Willems and Cavagna (1998) analyzed characteristics of running in children between the ages of 2 and 16. Vertical stiffness, measured as acceleration of the center of mass divided by vertical center of mass displacement, remained constant from ages 2 to 12 years. Normalized vertical stiffness steadily decreased from ages 2 to 12. An increase in vertical stiffness properties from age 13 to 16 illustrates the importance of using children for a proper analysis of the jumping program developed by Oregon State University. Conclusions from prior research concerning kinetics and kinematics of drop landing from adult subjects should be applied cautiously to prepubescent children.

Stiffness

Many researchers have described movement characteristics of the human body in terms of stiffness using a model of a mass attached to a massless linear spring. The mass-spring model has been used to describe stiffness of the leg and total body in activities such as hopping, running and drop landing. However, the methods of calculating stiffness vary (Blickhan, 1989; Ferris, Louie and Farley, 1998; Farley et al., 1998; Farley et al., 1991; Ferris and Farley, 1997; McMahon and Cheng, 1990; Farley and Gonzelez, 1996; Dalleau et al., 1998; Schepens, Willems and Cavagna, 1998).

As a typical mechanical system, the stiffness of a spring in a simple linear mass-spring system is calculated as the ratio of force to displacement:

$$k = \frac{F}{x} \tag{1}$$

where k is the stiffness, F is the force acting on the mass-spring system and y is the displacement of the mass. Schepens, Willems and Cavagna (1998) plotted vertical acceleration of the center of mass versus vertical displacement of the center of mass using data collected from children running across a force plate. Stiffness was calculated using the slope of the portion of the graph containing only upward displacement of the center of mass, neglecting the acceleration data recorded at impact containing peaks resultant from heel strike. Using acceleration instead of force in a stiffness calculation is perfectly acceptable since the acceleration is equal to force divided by a constant mass. However, by neglecting to include the impact portion of the acceleration-displacement curve in stiffness calculations the authors are suggesting that the body acts nonlinear at impact.

Ferris and Farley (1997) calculated vertical stiffness during hopping as the ratio of the peak vertical force to the maximum displacement of the center of mass. Using a linear mass-spring model they are assuming that the peak vertical force occurred at exactly the same instant the center of mass was maximally displaced. Farley and Gonzalez (1996) found that the relationship between force and displacement of the leg and total body center of mass while running was only linear following the peak force at impact from heel strike. They calculated two types of stiffness: leg stiffness and vertical stiffness. Leg stiffness was calculated as the ratio of the resultant force at the ground to the displacement in the leg spring at the point of maximal compression. The leg spring is defined as the distance from the ground to the greater trochanter. Vertical stiffness was calculated as the ratio of the vertical ground reaction force to the vertical displacement of the center of mass at maximal compression. Leg stiffness calculations allow stiffness to be calculated as the leg changes angles through a stride while running or walking, whereas vertical stiffness only considers the stiffness in the vertical direction. Conveniently, during running, the point of maximal compression occurred at the same time for both the leg spring and center of mass. McMahon and Cheng (1990) state that, "...except for hopping in place, the stiffness of the leg is not the same thing as kvert (vertical stiffness)." Therefore one could infer that vertical stiffness and leg stiffness would be the same in activities where only vertical motion occurred such as a drop jump, counter movement jump, hopping or landing with no horizontal velocity. However, in pilot data analyzed from drop landings, peak force does not coincide with maximum center of mass displacement.

Dyhre-Poulsen, Simonsen and Voigt (1991) calculated stiffness in subjects performing drop landings from a height of 60 cm using first order finite differences (equation 2).

$$k = \frac{(F_b - F_a)}{(x_b - x_a)} \tag{2}$$

This method of calculating stiffness allowed these researchers to calculate an instantaneous stiffness rather then a global stiffness for the whole system. While not stated in their research dividing the force by the displacement as in equation (1) assumes they used a linear mass spring model to represent their jumpers. However, modeling each increment in force divided by displacement from a drop landing with a linear mass spring would produce many different spring properties throughout one jump. This method makes it very difficult to make any conclusions concerning how the body acts as a system during impact.

A simple linear mass spring model works well for repetitive activities such as running and hopping where the center of mass oscillates similarly to a sine wave. However, the simple linear mass spring model does not represent the center of mass characteristics of the human body during landing from a height since the center of mass does not have the same beginning and ending vertical position. The center of mass displacement observed during landing has characteristics more representative of a single degree of freedom linear mass-spring-damper model (Robinovitch, Hayes and McMahon, 1991; Minetti et al., 1998). Adding dampening to the linear mass spring model attempts to account for inelastic components of the human body and different beginning and ending vertical positions within each trial.

Using a mechanical model to fit vertical displacement data allows for calculations of stiffness, and damping that describe how the body responds to drop landings from initial contact with the ground to standing at rest. Using a mechanical model to fit ground reaction force data allows for calculations of stiffness, and damping properties that describe how the body responds at initial impact. Splitting the landing into an impact specific signal (force data) and a global response signal (C.O.M. data) allows for a more clear picture of what is actually happening and is of particular interest in bone research. Specifically, modeling the force data with a mechanical model will provide an estimate of mechanical characteristics of the body during a reflexive muscular response to landing.

Reflex and Fatigue Characteristics of Muscle

There is the possibility of having many differences in joint kinematics and kinetics during landings based on differences in muscle length and stiffness in the hip and knee extensors and the ankle plantar flexors. Muscle reflex response and eccentric force production may also differ depending on how muscle characteristics change after performing repeated drop landings (Avela and Komi, 1998).

During drop landings, Dyhre-Poulsen, Simonsen and Voigt (1991) used the Hoffmann reflex, measured using electromyography, to represent the stretch reflex in the soleus, tibialis anterior and the medial head of the gastrocnemius to determine whether high muscle stiffness during landing would prevent energy at impact from being absorbed. Instead, high muscle stiffness would result in a more elastic response of the muscles and tendons causing the system to bounce. Therefore, it was hypothesized that upon landing from a jump there must be a mechanism that allows low muscle stiffness simultaneous with muscular contraction. The Hoffmann reflex, representing the stretch reflex, was reported as being strongly inhibited while landing from a jump, allowing a majority of energy at impact to be absorbed. The high initial ground reaction force peaks seen in ground reaction force traces from landings are speculated to be a result of short range muscle stiffness. Theoretically the peaks do not occur over a long period of time due to the breaking of the cross bridges caused by the high velocity stretch of the muscle.

Garland and McComas (1990) reported a loss of voluntary torque production at the ankle associated with a reduction in voluntary EMG activity of the soleus. The Hoffmann reflex was used as a measure of soleus motor neuron excitability and was found to decrease with fatigue. Fatigue was induced using ischemia and tetanic stimulation. Voluntary muscular contraction may decrease with repeated muscle contractions induced by performing multiple drop landings. However, it is unclear whether the peak forces at impact due to short range muscle stiffness would change with fatigue and it is unknown whether 100 drop landings from a height of 61 cm would be enough to induce fatigue in voluntary muscular contractions. Any changes in reflexive and voluntary properties of muscle will result in changes in how forces from the ground will be transmitted to the rest of the body and, therefore must be considered in any analysis of how forces might change across multiple drop landings.

Methods of Calculating Joint Kinetics

Several methods have been employed to estimate the resultant forces at joints in the body. Accelerometers have been used to estimate hip joint reaction forces (Bogert, Read, and Nigg, 1996; Bogert, Read, and Nigg, 1999). This methodology is based on the assumption that the body segment to which the accelerometers are attached is rigid (Bogert, Read and Nigg, 1996). This method of hip force estimation was not reliable during the impact phase of running and underestimated hip joint reaction forces by 20% compared to a standard rigid body model. In addition, the accelerometer method was reported to be accurate only when a subject performing an activity had all of his/her weight on one leg as the accelerometer data were being recorded. As suggested by Bogert, Read, and Nigg (1996), the use of accelerometers in joint force calculations is ideal for real time analysis of hip joint forces and moments as well as for activities with low frequency characteristics. However, this method does not consider the internal muscular forces that contribute to the joint reaction force.

Inverse dynamics methods of calculating joint forces and moments typically use estimated average anthropometric properties of the human body and assume each joint is connected by a frictionless pin. Gruber et al. reported large inaccuracies in using a rigid body model when calculating joint reaction forces due to the lack of consideration for soft tissue movement. Modifying the rigid body model to include estimated soft tissue movement, Gruber et al. developed the wobbling mass model which uses estimated coefficients for frequency of soft tissue movement and damping characteristics of the soft tissue in joint kinetic calculations. When analyzing a computer model of a drop jump from 40 cm, the wobbling mass model calculated vertical knee joint reaction forces that were nearly 2000 N greater than that calculated using rigid body dynamics. The calculated peak vertical ground reaction force was nearly 7000 N (11 times Body Weight) using the wobbling mass model and 10,000 N (16 BW) using the rigid body model. In addition, the wobbling mass model calculated peak vertical hip joint reaction force of 2100 N (3.5 BW) whereas the rigid body dynamics model produced a peak force of –2800 N (-4.5 BW). The difference in the sign is explained as being due to the wobbling mass model's inclusion of large soft tissue movement at impact. Both models consider the pure mechanics of the human body as linked segments and show hip joint reaction forces as being lower in magnitude than the ground reaction forces. Having hip joint reaction forces that are lower in magnitude than ground reaction forces is due to the fact that neither model considers muscle forces as contributing to the joint reaction forces.

Using a model that considers the action of 42 muscles in the lower extremities, Röhrle et al. (1984) reported peak vertical hip joint reaction forces 4.5 times greater than vertical ground reaction forces in walking at 1.2 m/s. In 1997, Bassey et al. compared ground reaction forces to compressive axial forces measured in an instrumented femoral implant. The forces in the implant, measured in jogging and jumping activities, were 1.5 – 3 times greater then the corresponding vertical ground reaction forces. Slow jumping with distinct take off and landing peaks provided the greatest forces in the hip implant. The average jump height was 5.8 cm. Lying down, the subject with the instrumented implant had compressive hip forces ranging from 200 – 300 N due to resultant muscle tension. An important finding reported in this research is the apparent linear relationship between ground reaction forces and hip joint compressive forces despite having a joint capsule comprised of many nonlinear materials such as muscles, ligaments and articular cartilage. Unfortunately the author's did not report any statistical values for their comparison of ground reaction forces to hip joint reaction forces.

Resultant hip forces from an 82 year old man with an instrumented femoral implant were reported to be nearly 3 times body weight while walking at 0.83 m/s and 4.7 times body weight while jogging at 1.6 m/s (Bergmann et al., 1995). Resultant hip forces from a 69 year old woman with an instrumented femoral implant were reported to be 4.7 times body weight while walking at 0.83 m/s and 8.7 times body weight when stumbling.

An 82 year old man with an instrumented femoral implant in each hip had forces ranging from 5.6-5.8 times body weight during running at 1.9 m/s and 7.2 times body weight when stumbling (Bergmann et al., 1993). Although ground reaction forces were not measured by Bergmann et al. (1993) and Bergmann et al. (1995) these values for hip joint forces are greater then typical ground reaction forces for walking and running (Munro, Miller and Fuglevand, 1987).

Joint reaction forces calculated using simplistic rigid body models should be interpreted as minimal estimates of joint forces. Measurements made using instrumented implants appear to provide the most accurate measures of hip joint reaction forces. However, during hip implant operations muscle site attachments are not always the same as they were on the original femur making the forces in the implant different then they would be with a completely healthy femur. During a drop landing activity, realistic hip joint reaction forces should be greater than measured ground reaction forces due to internal muscular forces, unfortunately a precise relationship between hip joint reaction forces and ground reaction forces is not yet known.

Before exercise programs can be implemented for osteoporosis prevention force characteristics resultant from the exercise programs must first be known. Comparing the known kinetics of an exercise program to force characteristics that have been associated with osteogenesis is an efficient method for developing exercise protocols designed to prevent osteoporosis.