AN ABSTRACT OF THE DISSERTATION OF

Jeremy J. Bauer for the degree of Doctor of Philosophy in Exercise and Sport Science presented on April 27, 2006.

Title: Defining Intensity of Skeletal Loading in Children.

Abstract Approved:

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Wilson C. Hayes   Christine M. Snow

While exercise can be prescribed for improving cardiovascular and muscle health, no prescription exists for increasing bone mass. Because bone deformation rate has been identified as an important variable related to osteogenesis, estimates of skeletal loading during human activities likely characterize the associated osteogenic stimulus. However, estimates of body segment parameters (BSPs) are needed to calculate skeletal loading. The preferred equations for calculating pediatric BSPs are based on 12 boys and have not been validated. To validate these equations for girls, we investigated whether equation-estimated BSPs differ from those derived using magnetic resonance imaging (MRI) and whether such differences cause differences in calculated joint kinetics during walking, running, and drop landings from three heights. We further compared hip joint kinetics among activities and to those at the ground. Left leg BSPs were estimated from MRI and using the equations in 10 girls. Joint kinetics were also calculated for each activity from recorded kinematics and ground reaction forces. With the exception of two shank variables, BSPs differed between methods. However, while these differences resulted in

statistically significant differences in joint kinetics for all activities, the differences were not sufficiently large to be of practical significance. Thus, equation-estimated BSPs appear suitable for use with girls. Significant relationships were found between peak forces and loading rates at the ground and hip, indicating that resultant hip loads can be predicted using forces at the ground. Walking and landings from 61cm had the lowest and highest forces, respectively. Forces during drop landings increased as height increased. Peak forces during running were not different than those for landings from 30 and 46cm. Loading rates at the ground during walking were less than for other activities, while those during running were less than for drop landings. There were no differences in loading rates among drop landings. Drop landings appear to have the characteristics most likely to cause osteogenesis. By quantifying ground forces and loading rates, we have provided a simple method for quantifying forces at the hip, a necessary step toward a better understanding of the relationship between loading and changes in bone mass at this site.

DEFINING INTENSITY OF SKELETAL LOADING IN CHILDREN

by
Jeremy J. Bauer

A DISSERTATION

submitted to

Oregon State University

in partial fulfillment of
the requirements for the
degree of

Doctor of Philosophy

Presented April 27, 2006
Commencement June 2006

Doctor of Philosophy dissertation of <u>Jeremy J. Bauer</u> presented on <u>April 27, 2006</u>.

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Chair of the Department of Nutrition and Exercise Sciences

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Dean of the Graduate School

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

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Jeremy J. Bauer, Author

ACKNOWLEDGEMENTS

As of the completion of this dissertation, I have been at Oregon State for 8.5 years. I received my M.S. in 2000 then proceeded to spend the following 6 years working on this dissertation. The most important lesson I've learned while I've been in Corvallis is that school is not as much about what you learn in class, but the experiences you have and the people you meet while in school. The need for balance in life is necessary to maintain sanity. Fortunately, to maintain that balance, I've been lucky enough to have the support of some truly amazing people.

Projects like this dissertation are never the result of only one person's work. From a student's point of view, having a committee that not only advises on research, but also cares a great deal about the student's work and well-being outside of the work is unreal. I've been fortunate to have such a committee. My committee consists of Wilson C. "Toby" Hayes, Christine M. Snow, Michael J. Pavol, Mark F. Costello and Elizabeth Sulzman. First and foremost, I'd like to thank Christine for both bringing me into the fascinating world of bone research and being such an amazing mentor. Despite being extremely busy running the OSU Bone Research Laboratory, teaching courses, and raising a family, Christine would hold individual weekly meetings with her students and not only discuss current projects, but ask about life in general with an unmistakably genuine concern. In addition, through Christine, I was introduced to another amazing mentor, Toby Hayes. In addition to having Toby as my major professor, I've also worked with/for him over the last 6 years as well. Toby repeatedly demonstrates that one can be a productive researcher and successful businessman while maintaining personal

relationships and prioritizing family life. Toby really is the busiest person I've ever

known, yet I'm always amazed at how he does not expect others around him to put the

same pressures on themselves. The respect Toby shows for his students and the people

that work for him instill a confidence in them that make them want to work hard for him.

What an incredible example of a great mentor. During the latter part of my M.S. Toby

introduced me to my minor professor in Mechanical Engineering, Mark Costello. In the

first few years of my dissertation, I spent many hours in Mark's office nearly pulling my

hair out while he calmly and patiently took me through example after example of

problems in dynamics relevant to my research. I now have a greater understanding of

engineering dynamics than I ever thought possible and use that knowledge everyday.

Mark is an amazing teacher. His current and future students are quite lucky. Mike Pavol

joined my committee just after my dissertation proposal and was instrumental to my

understanding of the motion system and programming language used in the research. On

top of that, Mike's attention to detail during both the research and writing process shows

exactly how feedback from others is always beneficial to improving the quality of the

work. I greatly appreciated the fact that, while Mike rarely seemed to leave campus due

to an extremely high work load, he always made time to relieve a little stress by coming

out to see our band (AMADAN) play. Mike is yet another example of a great mentor.

Elizabeth Sulzman, my graduate representative, is a person of incredible generosity.

While graduate students are trying to finish their dissertation, meeting all of the deadlines

set forth by the university, there are a million other little things that need to be taken care

of, such as adding a graduate representative to one's committee at the last minute.

Despite being over committed herself, Elizabeth agreed to serve on my committee and

proceeded to contribute greatly to the defense. If I had been in Crop and Soil Sciences, I would have most definitely wanted Elizabeth on my team. Finally, I'd like to thank two past members of my committee, Jeff Woldstad and Dan Williams. Jeff served as my graduate representative for both my proposal and my comprehensive exams before leaving for Texas. Dan was instrumental in making sure that my knowledge in statistics and research design was dialed in and properly applied to my research before leaving for Utah. I am extremely lucky to have had such an amazing committee. I am even luckier to have supportive friends and family.

I am extremely proud to come from a supportive, hard-working family. Being raised by parents that have supported EVERYTHING I've tried has taught me that there really are no limits to what one can accomplish. Their continual support of my education, not only in school, but in life continues to motivate me to work harder. I am very proud of my parents for managing to excel in what must be THE most difficult job in the world…raising children. I believe much of the credit for great parents can be attributed to amazing grandparents. Like my parents, my grandparents have always been in full support of everything I pursue…going as far as closing down bars (staying out late) while supporting my need to perform and play music. I only hope that one day I can play the same important role in someone else's life that my parents and grandparents have played in mine.

In addition to great parents and grandparents, my two brothers and their families have been incredibly supportive of me. Huge thanks to Jason for always saying, "Get that thing (dissertation) done!" and for keeping me supplied with functioning bicycles. A bicycle Jason built for me has lasted through 9 years of commuting in the Oregon rain,

bike races and many late nights of just pedaling around town to clear my head. As an added benefit, any time my sister-in-law, Heather, and their son Zachary made it to Corvallis with Jason was great. I was always left with great new bits of art to hang. Additionally, having the support of my younger brother Joel, his lovely wife, Debbie and their two children, Copen and Clayton is amazing. I love being an uncle and can't wait to have the time to play a more active role.

While one expects family to be supportive, an added bonus to life is that there are many great people out there willing to invest in friendships that blur the boundaries between friends and family. Filling the role of my sister, Kathy Gunter has been there for me from the day I first set foot in Corvallis. Well, "been there for me" really doesn't do her any justice. She dragged me into the Corvallis scene right away and made me feel very comfortable…very important when moving away from home. I'm very proud to have her as my friend and suuuper excited to get to teach baby Mia to snowboard and play music in a few years. Next, special thanks for tolerance of my busy schedule and endless schooling go to Jenn Ferris and her cat 7. Jenn put up with six years of my busy schedule and endless schooling and has been a pillar of support through it all. Also, yes, I'm acknowledging a cat in my dissertation, but sometimes a cat on the lap while typing late at night is just what is needed to keep the stress to a minimum. I need to acknowledge one of my best friends in the world, Kristen Westenskow (Proctor) for, always caring about how I am doing here in Oregon and for always making the time to come and see me in Oregon despite traveling long distances to do so. Also, Chad Harris deserves special recognition for being the reason I came to OSU in the first place. He has always been there to consult with as a lifelong committee member and friend and is still

always looking out for me. He was the first one to call me after my dissertation defense…truly amazing guy. I would like to thank my pal, Robyn Fuchs, for her continual support from afar and for the lovely post-defense daisies she sent and Kim Hannigan-Downs for the yummy ginger cookies through-out the years…mmm.

My dissertation did take a while longer to complete than I expected, but the distractions were well worth it. One of the first distractions came in the form of a great job in Injury Biomechanics at Hayes & Associates, Inc. One might not often see an acknowledgement of a workplace in a dissertation, but the people at Hayes & Associates are just absolutely exceptional. When half of my dissertation defense crowd was comprised of co-workers and my email INBOX was full of congrats messages from co-workers, I became acutely aware that the support and encouragement I receive on a daily basis from everyone at Hayes & Associates was very real. I'm fortunate to have a really great job working with amazing people. An additional distraction came in another form of work…music.

When I first started my program I was fortunate enough to fall into the company of a group of guys that have taken me places I never would have imagined. Eric Tonsfeldt, Andy Gross, Naoyuki Ochiai, Kevin Pardew, Mike Morrow and John Coleman, all current or past members of AMADAN, have shown me friendship and support that I never could have dreamed of having. Not only do I get to perform in from of hundreds of appreciative people 90+ times a year with these guys, but they all maintain a perspective on life very supportive of each other's interests outside of music, making for incredibly rewarding friendships. The AMADAN fans deserve special thanks in this acknowledgements section as well. While the stereotype of bar crowds usually

borders on drunken crazy people, I've been blown away by the number of people that are constantly interested to hear about my dissertation or how I'm progressing in school. In fact, I was actually so tired of the "when are you going to finish school" question from fans that a large part of the push to finish my degree came from fans of AMADAN. I'm very lucky to have the most perfect stress release outlet in the world…playing music with my friends to my friends. Fortunately, that stress relief activity led me to one of the most amazing people on this planet, Kelly Griffith. While I have only known Kelly for a brief time, I feel like I've known her forever. Her huge heart and loving support in the form of hugs, cards and care packages during the last days of my dissertation were unrivaled. I am extremely honored to have her love. I can't wait to reciprocate during her journey through medical school and every other aspect of life.

Lastly, I would like to thank the International Society of Biomechanics for partially funding my research and the Oregon State University Department of Nutrition and Exercise Sciences for supporting a majority of my education. As was true during completion of my M.S., all of the faculty in the department have been very supportive of me and are all very personal people. It has been a privilege to study at Oregon State University.

CONTRIBUTION OF AUTHORS

Dr. Hayes and Dr. Snow were instrumental in the conceptualization of all research presented in this dissertation. In addition, each contributed significantly to the editing of each manuscript. Dr. Hayes was instrumental in making the connections that allowed for the MRI data to be collected for this research. Dr. Pavol contributed significantly to the development of data collection methods such as writing and editing the code used in calculated joint kinetics. In addition, Dr. Pavol invested a considerable amount of time to the editing of each manuscript.

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DEDICATION

This dissertation is dedicated to my hard working parents and grandparents.

INTRODUCTION

Osteoporosis is a disease of crisis proportions in the United States, associated with more than 700,000 vertebral and 300,000 hip fractures annually (National Osteoporosis Foundation, 2002). The National Institutes of Health (NIH) has emphasized that increasing peak bone mass during youth is essential to preventing fractures later in life . However, there is currently no clear prescription for increasing peak bone mass in youth

A majority of human-based research related to bone and exercise has used broad programs of multiple resistance training exercises and cardiovascular regimens to increase bone mass (Heinonen, et al., 1998; Kerr, et al., 2001; Kerr, et al., 1996; Kohrt, et al., 1997; Kohrt, et al., 1995; Lohman, et al., 1995; Nelson, et al., 1994; Pruitt, et al., 1992). However, these broad programs do not allow clinicians to develop a relationship between the loading stimulus and changes in bone mass since the intensity component of the exercise does not directly relate to bone loading. Additionally, since multiple forms of loading are used, it is nearly impossible to associate one type of loading with changes in bone. The justifications for the modes of exercise used to date have been based on muscle- and cardiovascular-based programs. For example, the intensity metric for the cardiovascular system and skeletal muscle is a percentage of maximum heart rate or maximum VO_2 and a percentage of 1 repetition maximum (1 RM) or 6-8 RM, respectively. The intensity metric for bone in humans has yet to be defined. Fortunately, animal studies have provided a wealth of information through randomized controlled, trials concerning variables associated with osteogenesis.

Using an animal model, strain rate (rate of bone deformation) has been identified as the key variable related to the osteogenic response of bone (Turner, 1998). Strain rate is a function of the magnitude of strain (bone deformation) and the frequency of loading. However, strain is measured directly from the bone in animals, and cannot be ethically measured in humans. Therefore, a metric based on external forces that cause bone to deform and their rate of application has been proposed previously (Bauer, et al., 2001; Turner and Robling, 2003), but has not been widely used. Thus, in order to better derive exercise regimens aimed at optimizing hip bone mass and preventing fractures, it is important to be able to quantify a measurable stimulus (force or loading rate) related to bone loading.

The purpose of this research is to describe and compare several common childhood activities in terms relating to bone loading. Currently, the standard method used to estimate loading at the hip joint, inverse dynamics, requires some knowledge of the ground reaction forces, the motion of the limbs and the body segment parameters associated with each subject. Accurate determination of joint reaction forces and moments require an accurate estimation of the body segment parameters and the location of joint centers. While body segment parameters have been physically measured in adult cadavers, these parameters for children have been estimated using photogrammetry, but not validated. Therefore, we asked: 1) Do pediatric body segment parameters developed from MRI differ from body segment parameters derived from photogrammetry-based regression equations? If so, 2) Do the differences in body segment parameters result in significant differences in inverse dynamics-based joint kinetics at the hip and knee during normal gait, running and drop landing? And 3) What activity has loading characteristics

most closely resembling the loading patterns shown to increase bone mass in the animal model? What follows are three manuscripts, each designed to address issues related to quantifying skeletal loading in children.

Chapter 2 addresses the initial assumptions made when estimating skeletal loads in children during normal gait. In order to estimate key joint force variables such as joint forces, loading rates, moments and powers, one needs to know several important characteristics of a subject's segments, including: mass, center-of-mass location, and mass moment of inertia. While published regression equations designed to predict these parameters in children exist, the equations were based on a population of only 12 boys and not validated in predicting segment inertia. Therefore, one cannot assume that the regression equations are generalizable to a more diverse population of children.

While Chapter 2 addresses the effects of body segment parameter assumptions in normal gait, the goal of Chapter 3 is to investigate whether these assumptions in body segment parameters hold true when comparing joint kinetics from activities outside the realm of normal gait such as running or landing from a height. Chapters 2 and 3 help answer the questions as to whether or not existing regression equations aimed at predicting body segment parameters in children are valid.

Chapter 4 is then designed to use the knowledge gained from Chapters 2 and 3 in comparing skeletal loading among the three activities previously investigated: 1) walking, 2) running, and 3) drop landings from 30, 46 and 61 cm. The goal of Chapter 4 is two-fold: 1) To identify an intensity variable that will allow clinicians and researchers to describe activities in terms relating to skeletal loading and 2) to compare the loading

patterns of the aforementioned activities as they relate to *in-vivo* bone loading variables

that have been reported in the animal-based literature.

MRI DERIVED BODY SEGMENT PARAMETERS OF CHILDREN DIFFER FROM AGE-BASED ESTIMATES DERIVED USING PHOTOGRAMMETRY

Jeremy J. Bauer, Michael J. Pavol, Christine M. Snow and Wilson C. Hayes

2.1 Abstract (Word count: 239)

Body segment parameters are required when researching joint kinetics using inverse dynamics models. However, the only regression equations for estimating pediatric body segment parameters across a wide age range were based on 12 boys and developed using photogrammetry, which has not been validated for predicting segment inertia (Jensen, 1986). To validate these equations for use in girls, we asked whether body segment parameters estimated from these equations differ from parameters measured using MRI. If so, do the differences cause significant differences in joint kinetics during normal gait? Body segment parameters were estimated from axial MRIs of the left thigh and shank of 10 healthy girls (9.6 ± 0.9 yrs) and compared to those from Jensen's equations. Kinematics and kinetics were collected for 10 walking trials. Extrema in hip and knee moments and powers were compared between the two sets of body segment parameters. With the exception of the shank mass center and radius of gyration, body segment parameters measured using MRI were significantly different than those estimated using regression equations. The differences in body segment parameters resulted in significant differences in sagittal-plane joint moments and powers during gait. Nevertheless, it is doubtful that even the greatest differences in kinetics are practically significant (0.3%BW$\times$BH and 0.7%BW$\times$BH/s for moments and power, respectively). Therefore, body segment parameters estimated using Jensen's regression equations are a suitable substitute for more detailed anatomical imaging of 8-10 year old girls when quantifying joint kinetics during gait.

2.2 Introduction (Word count: 3082)

Body segment parameters such as mass, mass moments of inertia and mass center location are required when calculating joint kinetics from inverse dynamics models or forward dynamics simulations. For such models, the validity of the predictions is dependent on the validity of the input data. When studying adults, there are many corresponding data sources upon which to base estimates of body segment parameters (Chandler, et al., 1975; Clauser, et al., 1969; Dempster, 1955; Zatsiorsky and Seluyanov, 1983). However, these parameter estimates from adults are not valid for use with a pediatric population, as there is a redistribution of mass among the segments during development (Jensen, 1989).

To date, published sources of pediatric body segment parameters have employed either photogrammetry (Jensen, 1978; Jensen, 1986; Jensen, 1989; Jensen and Nassas, 1988; Yokoi, et al., 1986), photogrammetry combined with anthropometric measures (Ackland, et al., 1988), or Dual-Energy X-ray Absorptiometry (DXA)(Ganley and Powers, 2004a). Photogrammetry makes use of photographs of the body in the frontal and sagittal planes to define elliptical slices, spaced along the length of each segment. From these sections, assumed segment densities are used to estimate mass, mass moments of inertia, and mass center location. In contrast, DXA-based methods directly estimate these segment densities from a scan of the body in the frontal plane by a bone densitometer. Notable limitations of these methods are that photogrammetry has not been validated in estimating segment moments of inertia (Yokoi, et al., 1986), whereas DXA employs ionizing radiation.

An alternative method for determining body segment parameters in living humans is magnetic resonance imaging (MRI) (Cheng, et al., 2000; Martin, et al., 1989; Mungiole and Martin, 1990; Pearsall, et al., 1996; Pearsall, et al., 1994). MRI allows the specific determination of tissue types and distributions within slices along the length of a segment, without the health risks associated with ionizing radiation, and has been validated for estimating segment mass, mass moment of inertia, and mass center location. Despite this, MRI has not been applied to date in determining pediatric body segment parameters.

Each of the aforementioned methods provides subject-specific body segment parameters that can be applied to a model of the subject for the calculation of joint kinetics. Compared to parameters estimated using generalized prediction equations, subject-specific body segment parameters reduce error in the model input (Durkin and Dowling, 2003) and can thus reduce the error in the output of an individual's joint kinetic calculations. However, the time, expense, and access to specialized equipment needed to develop subject-specific parameters precipitates the need for a simple, robust method for predicting body segment parameters in children.

Currently, regression equations provide the most convenient, age-specific method of estimating body segment parameters in pediatric populations. However, the only equations available for estimating pediatric body segment parameters across a wide age range are based on just 12 boys, and have not been validated in predicting mass moments of inertia and mass center location (Jensen, 1986; Jensen, 1989). Additionally, the effects of using Jensen's generalized, age-based regression estimates of body segment parameters, rather than detailed subject-specific parameters, on computed joint kinetics are unknown. Since joint kinetic calculations are often the basis for surgical decisions

involving the correction of pathologic gait in children, an investigation into the validity of existing body segment parameter estimation methods is warranted.

In an attempt to validate Jensen's regression equations for use in girls, we sought to answer the following questions: Do body segment parameters estimated for the thigh and shank from Jensen's (1986) photogrammetry-based regression equations based on boys differ from subject-specific segment parameters measured using MRI in girls? If so, do the differences result in significant differences in inverse dynamics-based sagittal-plane hip and knee moments and powers during normal gait? We hypothesized that body segment parameters acquired from MRI would be sufficiently different from those based on regression equations to cause significant differences in computed joint kinetics.

2.3 Methods

Subjects

Ten girls (age: 9.6 ± 0.9 yrs; mass: 37.0 ± 7.1 kg, height: 141.0 ± 9.4 cm, BMI: 18.5 ± 1.8; mean $\pm$ SD) were recruited. Each subject and her parent or guardian provided written informed consent to participate. Institutional Review Board approval was obtained.

MRI Body Segment Parameters

The left lower extremity of each subject was imaged using a GE Signa 1.5T MRI machine (GE Healthcare Technologies, Waukesha, WI, USA). A series of 6 mm-thick slices, spaced 8 mm apart along the longitudinal axis of the leg, were imaged using a T1 weighted spin-echo sequence with an 850 ms repetition time, 8 ms echo time, and a 165 x

220 mm field of view. Slices were imaged from proximal to the femoral head through the calcaneus. Due to the coil length of 48 cm, three passes were required to image an entire leg: 1) proximal 2/3 of the thigh, 2) distal 1/3 of the thigh and proximal 1/3 of the shank, and 3) distal 2/3 of the shank. Subjects were not repositioned between passes.

Segmentation of the leg into thigh and shank segments followed the boundaries employed by Jensen (Jensen, 1978). The thigh was sectioned proximally along a line from the crease in the crotch through the hip joint center and distally through the knee joint center (Clauser, et al., 1969). The shank was sectioned proximally through the knee joint center and distally through the ankle joint center.

In each MRI slice, the different tissue types (i.e. muscle, bone, fat) were identified based on pixel brightness (Figure 1) using Photoshop 7.0 (Adobe Systems, San Jose, CA, USA). Tissue area was quantified by summing the pixels and multiplying by the pixel area (= 0.25 mm^2). The scaling from pixels to millimeters was determined from fiduciary marks on the images. For each tissue type, tissue volume within a slice was assumed to be the frustum of a cone and quantified as,

$$V_i = (A_i + A_{i+1})/2 \text{ x } h \tag{1}$$

where V_i is the volume of slice i, A_i and A_{i+1} are the tissue areas in adjacent scans, and h is the distance between slices (= 8 mm). To obtain the mass of each slice, volumes for each tissue were multiplied by their respective density (cortical bone = 1.705 g/cm^3, trabecular bone = 1.1 g/cm^3, muscle = 1.067 g/cm^3, and fat = 0.96 g/cm^3) and summed (Martin, et al., 1989; Mungiole and Martin, 1990). The center of mass location from the proximal end of a segment (COM_p) was calculated as,

$$COM_p = \left(\sum m_i x_i \right) / M \tag{2}$$

where m_i is the mass of slice i, x_i is the distance from the center of slice i to the proximal end of the segment, and M is the mass of the segment. The mass moment of inertia about a transverse axis through the segment center of mass (I_{CM}) was calculated as,

$$I_{CM} = \sum \left(\frac{1}{12} m_i h^2 + \frac{1}{4} m_i r_i^2 + m_i CM_i^2 \right) \tag{3}$$

where r_i is the average radius of cylindrical slice i and CM_i is the distance from the center of slice i to the segment center of mass. From the parallel axis theorem, the moment of inertia about the proximal end of the segment (I_p) was,

$$I_p = I_{CM} + M * COM_p \tag{4}$$

and the corresponding radius of gyration (ROG_p) was,

$$ROG_p = (I_p/M)^{1/2} \tag{5}$$

The radius of gyration about the longitudinal axis was similarly determined. Center of mass locations and radii of gyration were expressed as percentages of the segment length.

Measurement validity was assessed using one intact bovine shank containing fat, muscle, and bone. The shank and a piece of plexiglass of matching length and width (30.5 x 13.7 x 0.5 cm) were weighed using a digital scale. The shank was placed on the plexiglass and this system was balanced on a fulcrum to locate its center of mass. Knowing the masses and centers of mass of the plexiglass and shank/plexiglass system allowed the determination of the shank center of mass. The MRI-derived shank mass was 3.7% (0.12 kg) less than the measured mass. The MRI-derived center of mass was computed to be 1.5% of the shank length (5.5 mm) more distal than the actual center of mass.

MRI Hip Joint Center Location

Coronal MRI slices of the pelvis were imaged at 4 mm intervals. The slice

through the centers of the femoral heads and the most anterior slice in which the left

anterior superior iliac spine (ASIS) was visible were identified and exported into

Photoshop for measurement. Pelvic width was calculated as the distance between the left

and right ASIS. Because the right ASIS did not appear in the MRIs, its location was

computed assuming bilateral symmetry about the midpoint between the hip joint centers.

The displacement from the left ASIS to the left hip joint center was expressed in each

anatomical direction as a percentage of the calculated pelvic width (21.5 ± 2.9% Posterior,

14.5 ± 2.6% Medial, 32.8 ± 4.9% Inferior; mean ± SD).

Gait Analysis

Each subject performed trials of walking approximately 20 m across the

laboratory. The first 10 trials in which the subject successfully contacted a force platform

(Bertec, Columbus, OH, USA) located at the midpoint of the gait path with her left foot

were analyzed. Subjects were not made aware that they were to step on the force

platform. They were instructed to walk at a self-selected pace, maintaining eye contact

with a target positioned at head level at the opposite end of the laboratory. The average

walking speed, as determined from the sacrum marker, was 1.5 ± 0.1 m/s across subjects.

A 6-camera motion capture system (Vicon Peak, Lake Forest, CA, USA) sampled

the positions of 20 reflective markers, attached to the lower extremities and pelvis, at 120

Hz. Ground reaction forces were sampled at 1080 Hz and synchronized with the motion

capture data. Before processing, marker data were interpolated to the output frequency of

540 Hz using cubic splines, and filtered with a 4th-order Butterworth no-lag filter at a cutoff frequency of 10 Hz, as determined by residual analysis (Winter, 1990).

Sagittal-plane knee and hip angles, moments, and powers were calculated through a three-dimensional, inverse-dynamics analysis (*BodyBuilder*, Vicon Peak). Joint center locations and body segment orientations were computed from the marker data using transformations derived from anthropometric measurements and an initial trial of quiet standing. Hip joint center locations were determined using subject-specific MRI data in conjunction with the ASIS markers. Joint rotations and moments about the mediolateral axis of the proximal segment were extracted for analysis, with power computed as the product of the moment and angular velocity.

Each trial was processed twice, once using body segment parameters estimated from the pediatric regression equations of Jensen (1986) and once using subject-specific thigh and shank parameters (i.e. mass, mass center location, mass moments of inertia) derived from the MRI data. Key extrema in the time histories of the knee and hip sagittal-plane moments and powers during stance were identified and averaged across trials for each subject (Figure 2). Moments and powers were normalized to body height (BH) and body weight (BW).

Statistics

Paired t-tests were used to assess differences in segment mass, mass center location, and radius of gyration between the regression and MRI methods. Paired t-tests were also used to assess differences in the gait kinetic variables between methods. Differences were considered statistically significant at p-values less than 0.01.

2.4 Results

With the exception of the shank mass center location and radius of gyration, body segment parameters measured using MRI were significantly different than those estimated using regression equations (Table 1). Thigh mass derived from MRI was 7.3% greater than that derived from regression equations, whereas shank mass derived from MRI was 26.7% less than that derived from regression equations. MRI-derived estimates of the thigh center of mass location and of the transverse-axis radius of gyration about the hip were 5.9 ± 0.7 % and 3.8 ± 0.1 % of segment length more proximal, respectively, compared to regression equation predictions. Estimates of the shank center of mass location and transverse-axis radius of gyration about the knee did not differ between methods.

The differences in thigh and shank segment parameters resulted in statistically significant differences in sagittal-plane hip and knee moments and powers during gait. Significant differences were found in 3 of 5 moment extrema and in 4 of 7 power extrema (Tables 2 & 3). These differences were small, however, with maximum differences of 0.3 ± 0.1 %BW×BH and 0.7 ± 0.3 %BW×BH/s for moments and power, respectively.

2.5 Discussion

The purpose of this research was to determine whether Jensen's (1986) regression equations, developed from a population of boys and not yet validated in predicting segment mass center location and mass moments of inertia, were valid for predicting body segment parameters in girls. In addition, we sought to determine whether errors associated with applying Jensen's regression equations to girls would produce significant

differences in joint kinetics during gait, since these kinetics are often bases for clinical decisions to treat abnormal gait. We found that, with a few exceptions, Jensen's equations were not valid in predicting body segment parameters in our population of girls. This lack of validation was evidenced by the greater thigh mass, lesser shank mass, more proximal thigh center of mass, and smaller thigh transverse-axis radius of gyration measured using MRI compared to regression equation estimates. The differences in body segment parameters resulted in many statistically significant differences in joint moments and powers during gait.

Jensen's regression equations are currently the most convenient means of estimating body segment parameters in children across a wide age range. However, their validity appears to be limited. Arguably, a primary limitation is the use of chronological age as the sole predictor variable. Because children have differing body types and develop at differing rates, children within the same chronological age group will have different body segment parameter proportions. Indeed, Yokoi et al. (1986) found that body segment proportions varied by age, sex, and Kaup's index ($[mass(g)]/[height(cm)]^2$) in 255 children ranging from 3-15 years old. To address this issue of chronological age versus skeletal maturity, Ackland et al. (1988) developed a set of equations for predicting pediatric body segment parameters based on a few anthropometric variables. Such an ability to account for effects of anatomical differences on body segment parameters is potentially important in that the primary application of gait analysis in children is to pathological gait. Unfortunately, the equations of Ackland et al. were based on boys of similar age (13.7 ± 1.4 years), hence their applicability to a more diverse pediatric population is unknown. The remaining option, to obtain subject-specific body segment

parameters for each subject, can provide the greatest accuracy but is impractical for most laboratories and clinics due to the expense, time, and equipment needed. Thus, despite their present lack of validity, Jensen's regression equations arguably remain the most practical and widely applicable method for estimating body segment parameters across the pediatric age range. As such, it was important to assess the consequences of the errors associated with these equations.

The use of body segment parameters from Jensen's (1986) regression equations instead of MRI-derived, subject-specific parameters resulted in many statistically significant differences in joint moments and powers during gait. From a practical, clinical standpoint, however, and in looking at the moment and power time-histories (Figure 2), the differences in joint kinetics between methods were small. Therefore, while Jensen's equations were not valid in predicting all lower extremity body segment parameters, the lack of validity did not make a practically significant difference in joint kinetic calculations. Consistent with our findings, Ganley and Powers (2004a) found that body segment parameters estimated using adult cadaver-based equations differed from DXA-derived values in 7-13 year-old children; however, the differences did not result in significant root-mean-square errors between the associated joint moment time-histories calculated during normal gait. The effects of the differences in body segment parameters on the calculated joint moments appeared to be greater during the swing phase than during the stance phase of gait (Ganley and Powers, 2004b). Together, these studies and the present suggest that joint kinetic calculations during normal gait in children are relatively insensitive to sex- and age-differences in body segment parameters.

To our knowledge, this study is the first to investigate the validity of Jensen's (1986) regression equations in estimating pediatric body segment parameters for use in joint kinetic calculations in girls. Because we processed a large number of walking trials for each subject, our data were likely representative of each subject's normal gait. We also minimized the error in joint kinetics by measuring and applying subject-specific hip joint center locations to the rigid body model used in the kinetic calculations. Of consideration, however, is that only girls were tested. The extent to which the observed differences in body segment parameters and joint kinetics were specifically related to sex is therefore unknown. Our subjects were also very similar in age and body type. Thus, Jensen's regression equations were validated only for a small population.

On a more technical issue, an inherent limitation with MRI is that slice thickness can contribute to errors in mass, mass center location, and radius of gyration. Our validation sample had errors in mass and mass center location of 3.7% and 5.5 mm, respectively. Given the MRI slice spacing (6 mm thick + 2 mm space), the proximal and distal ends of the segment could be located at any point along the first and last 8 mm slices of a segment, respectively. For example, 310 mm of MRI slices were required to image the 305 mm length of the validation sample. The resulting 5 mm discrepancy in the segment length of our validation sample could affect the calculations of mass, mass center location, and radius of gyration. Given the inherent error, our body segment parameter estimates were well within the range reported in other studies (Table 4).

While statistically significant differences were found when comparing body segment parameters and joint kinetics at the knee and hip between parameter estimation methods, it is doubtful that even the greatest difference in joint kinetics calculated in this

study is practically significant. The greatest percent differences in joint kinetics amounted to a hip flexor moment (HM2) difference of only 0.3 %BW×BH and a hip concentric power (HP3) difference of only 0.7 %BW×HT/s or 3.7 Watts. Therefore, we are confident that body segment parameters estimated using Jensen's (1986) regression equations are a suitable substitute for more detailed anatomical imaging of normal 8-10 year old girls when quantifying joint kinetics during stance in normal gait. However, the development of body segment parameter prediction equations based on anthropometry and validated measurement methods, such as MRI, would be a significant contribution in pediatric biomechanics and should be pursued further.

2.6 Acknowledgements

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2.8 Figures
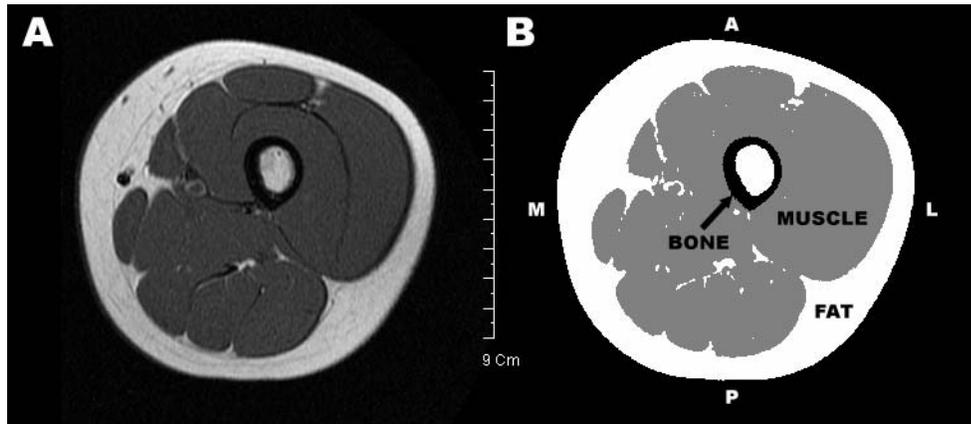
Figure 2.1. Inferior view of a representative slice through the distal third of the thigh. A)

Original MRI, B) Processed slice depicting anatomical orientation and tissue types. A =

anterior; L = lateral; P = posterior; M = medial.
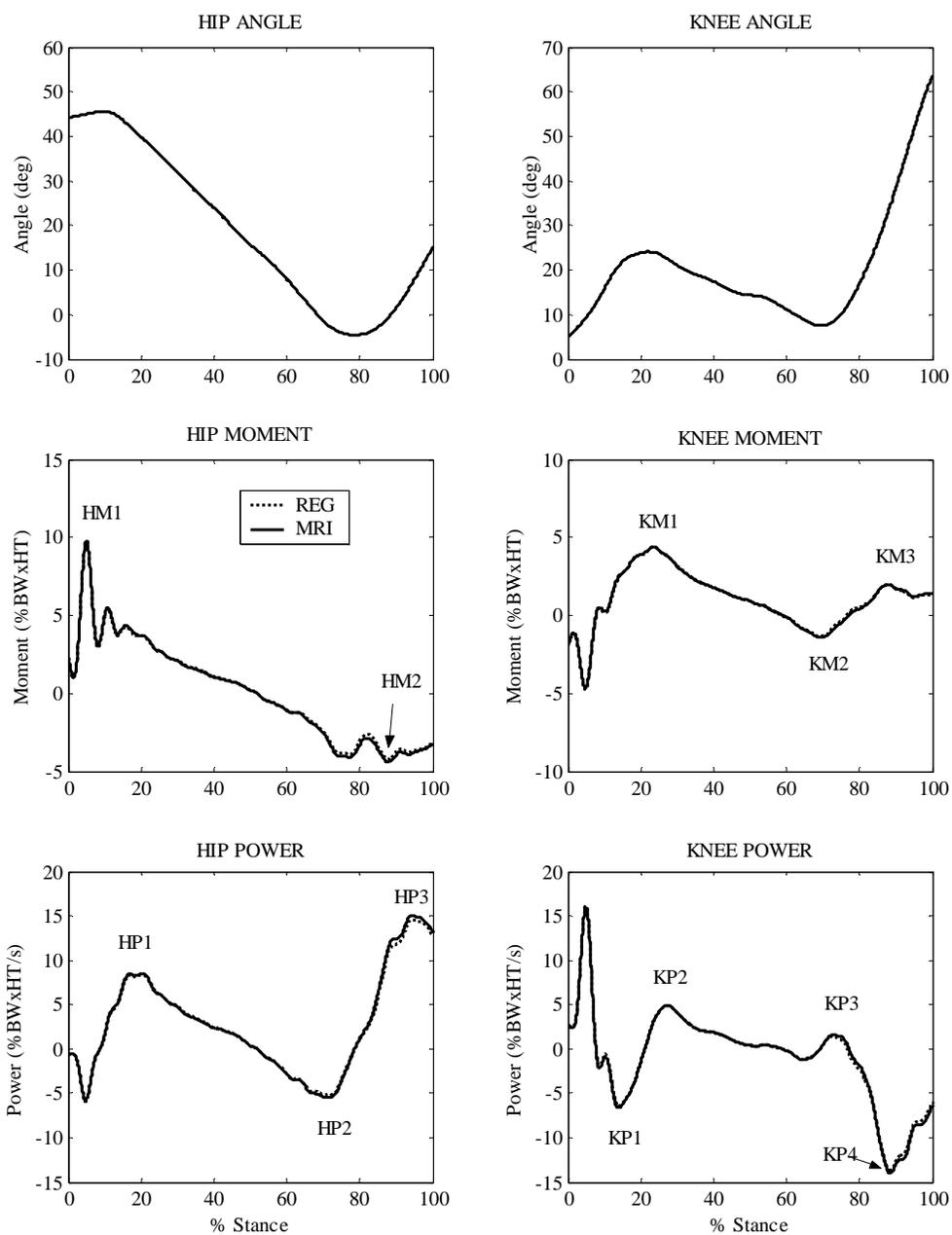
Figure 2.2. Representative data for hip and knee sagittal-plane kinematics and kinetics

during the stance phase of gait, showing extrema included in the analysis. Data are

shown for body segment parameters predicted by Jensen's (1986) regression equations

(REG; dotted lines) and derived from MRI (solid lines). Positive values represent flexion

angles, extension moments, and concentric power. Moments analyzed include: peak hip extension moment (HM1), peak hip flexion moment (HM2), peak knee extension moment in early stance (KM1), peak knee flexion moment (KM2), and peak knee extension moment during push-off (KM3). Powers analyzed include: peak hip concentric power during heel strike (HP1), peak hip eccentric power (HP2), peak hip concentric power during push-off (HP3), peak knee eccentric power during heel strike (KP1), first peak knee concentric power in mid-stance (KP2), second peak knee concentric power in mid-stance (KP3), and peak knee eccentric power during push-off (KP4) (McGibbon and Krebs, 2004). NOTE: Ankle data are not presented, as no body segment parameters were determined for the foot using MRI.

Table 2.1. Body segment parameter comparison between estimation methods.

	MRI	Regression	P-Value
Thigh mass (%BW)	10.9 ± 0.7	10.2 ± 0.3	0.001
Shank mass (%BW)	4.1 ± 0.3	5.0 ± 0.1	0.000
Thigh COM (%SEG LEN)	40.6 ± 0.8	46.5 ± 0.1	0.000
Shank COM (%SEG LEN)	41.5 ± 1.3	42.3 ± 0.3	0.049
Thigh ROG (%SEG LEN)	50.3 ± 0.6	54.1 ± 0.1	0.000
Shank ROG (%SEG LEN)	50.4 ± 0.9	50.9 ± 0.2	0.118

* Center of Mass (COM) and Radius of Gyration (ROG) reported from proximal end of

segment. Values are mean ± SD. BW = body weight; SEG LEN = segment length.

Table 2.2. Peak joint moment comparison between body segment parameter estimation methods.

Moment (%BW×HT)	MRI	Regression	P-Value
Hip Extension Moment (HM1)	10.13 ± 1.57	10.08 ± 1.58	0.259
Hip Flexion Moment (HM2)	-5.51 ± 1.27	-5.26 ± 1.26	0.000
Knee Extension Moment 1 (KM1)	5.25 ± 1.41	5.27 ± 1.40	0.146
Knee Flexion Moment (KM2)	-1.30 ± 0.55	-1.25 ± 0.56	0.001
Knee Extension Moment 2 (KM3)	2.41 ± 0.69	2.34 ± 0.70	0.000

Values are mean ± SD. BW = body weight; HT = height.

Table 2.3. Peak joint power comparison between body segment parameter estimation methods.

Power (%BW×HT/s)	MRI	Regression	P-Value
Hip Concentric Power 1 (HP1)	8.70 ± 2.50	8.53 ± 2.43	0.015
Hip Eccentric Power (HP2)	-5.83 ± 2.14	-5.71 ± 2.23	0.010
Hip Concentric Power 2 (HP3)	14.95 ± 5.10	14.24 ± 4.91	0.000
Knee Eccentric Power 1 (KP1)	-12.77 ± 6.24	-12.90 ± 6.23	0.005
Knee Concentric Power 1 (KP2)	6.30 ± 3.78	6.33 ± 3.77	0.044
Knee Concentric Power 2 (KP3)	1.43 ± 0.64	1.38 ± 0.63	0.060
Knee Eccentric Power 2 (KP4)	-13.89 ± 3.74	-13.55 ± 3.69	0.000

Values are mean ± SD. BW = body weight; HT = height.

Table 2.4. Comparison of MRI-derived body segment parameters with other studies.

Research	N	Age (yr)	Thigh Mass (%BM)	Shank Mass (%BM)	Thigh COM %[*]	Shank COM %[*]	Thigh ROG %[**]	Shank ROG %[**]
Present Study	*10*	*9.6*	*10.9*	*4.1*	*40.6*	*41.5*	*29.7*	*27.8*
Ganley & Powers (2004)	16	9-10	11.4	4.7	46.5	41.6	25.2	27.4
Ackland (1988)	63	13.7	12.8	5.4	43.6	41.8		
Yokoi (1986)	11	9-11	11.0	4.5	47.5	41.2		
Yokoi (1986)	6	9-11	10.7	4.2	47.3	40.3		
Jensen (1986)	12	9.6	10.2	5.0	46.5	42.3		

[*] % of segment length reported from proximal end of segment; [**] % of segment length reported from COM of segment; COM = center of mass; ROG = radius of gyration; BM = body mass.

DIFFERENCES IN SEGMENT PARAMETERS DO NOT AFFECT PEDIATRIC JOINT KINETICS IN RUNNING AND LANDING

Jeremy J. Bauer, Michael J. Pavol, Christine M. Snow and Wilson C. Hayes

3.1 Abstract

Joint kinematics and kinetics can influence surgical choices for correcting pathologic gait in children. Previously, in an attempt to validate pediatric body segment parameters (BSP) used to calculate joint kinetics, we found that BSPs measured using MRI in children were significantly different than those estimated using age-based regression equations. However, for quasi-static activities such as gait, these differences did not lead to practically significant differences in joint kinetics. To extend these BSP comparisons to more dynamic activities such as running and drop landings, we asked: Do the differences between BSPs derived from MRI and regression equations result in significant differences in joint kinetics during running and drop landing? Extrema in hip and knee moments and powers were compared between the two sets of BSPs within each activity. The differences in BSPs resulted in significant differences in sagittal-plane joint kinetics during both running and drop landing. Nevertheless, even the greatest differences in kinetics during running were not practically significant (0.3 ± 0.2 %BW*BH and 1.4 ± 0.8 %BW*BH/s for moments and power, respectively). However, while the greatest difference in kinetics during drop landings was small (0.6 ± 0.3 %BW*BH), the largest difference in power was more substantial (4.2 ± 3.1 %BW*BH/s). Therefore, BSPs estimated using age-based regression equations appear to be a suitable substitute for more detailed anatomical imaging of 8-10 year old girls when quantifying joint kinetics during running. However, if power is the main outcome variable in drop landing analyses, clinicians should be cautious.

3.2 Introduction

Joint kinetic calculations are often times the basis for surgical decisions involving the correction of pathologic gait in children (Gage, 1995; Smith, et al., 2004). The accuracy of these calculations is dependent on the quality of the kinematics, kinetics, equations of motion and body segment parameters used in the dynamic model. Well-established methods have been developed, validated and accepted for estimating joint kinetics from forces and motion sampled from adults and children performing various tasks (Kadaba, et al., 1989; Winter, 1990). However, while body segment parameters have been extensively quantified and validated in adults, these adult parameters have not been validated for use in children.

To date, equations and proportions used to predict body segment parameters in children have been developed using either photogrammetry (Jensen, 1978; Jensen, 1986; Jensen, 1989; Jensen and Nassas, 1988; Yokoi, et al., 1986), photogrammetry combined with anthropometric measures (Ackland, et al., 1988) or Dual-Energy X-ray Absorptiometry (Ganley and Powers, 2004a). While investigating the validity of Jensen's (1986) age-based regression equations, we found that, despite statistically significant differences between the MRI and regression based body segment parameters, differences in joint kinetics during normal gait were not practically significant in adolescent girls (Bauer et al., in review). However, while age-based regression equations appear to be acceptable for use in calculating joint kinetics in normal walking, the effects of varying body segment parameters when analyzing more dynamic activities has yet to be examined in children.

In an attempt to validate Jensen's (1986) age-based regression equations for use in girls performing activities other than walking, we asked: Do the differences in MRI derived body segment parameters and body segment parameters estimated using age-based regression equations result in significant differences in inverse dynamics based joint kinetics at the hip and knee during running and drop landing? Based on our earlier research in walking, we hypothesize that differences in body segment parameters will not lead to practically significant differences in inverse dynamics based joint kinetics.

3.3 Methods

Subjects

Ten girls (age: 9.6 ± 0.9 yrs; mass: 37.0 ± 7.1 kg, height: 141.0 ± 9.4 cm, BMI: 18.5 ± 1.8; mean $\pm$ SD) were recruited. Subjects provided written assent after reading an age-appropriate study description. Each subject's parent or guardian also provided written consent for their daughter to participate. The protocol was approved by the Oregon State University Institutional Review Board.

MRI Body Segment Parameters

These methods have been reported previously, so will be summarized only briefly (Bauer et al., in review). The left lower extremity of each subject was imaged using a GE Signa 1.5T MRI machine (LX Horizon, GE Healthcare Technologies, Waukesha, WI, USA). A series of 6 mm-thick slices spaced 8 mm apart along the longitudinal axis of the left leg were imaged using a T1 weighted spin-echo sequence. Slices were imaged from proximal to the femoral head through the calcaneus. Segmentation of the leg into the

thigh and shank segments followed the same boundaries employed by Jensen in deriving

body segment parameters using the photogrammetric method (Jensen, 1978). In each

MRI slice, image processing software (Adobe Photoshop 7.0, Adobe Systems, Inc., San

Jose, CA, USA) was used to identify the various tissue types (i.e. muscle, bone, fat)

based on pixel brightness (Figure 3.1). Tissue area was quantified by summing the pixels

and multiplying by the pixel area. For each tissue type, tissue volume within a slice was

assumed to be the frustum of a cone. To obtain the mass of each slice, volumes for each

tissue were multiplied by their respective densities (cortical bone = 1.705 g/cm^3,

trabecular bone = 1.1 g/cm^3, muscle = 1.067 g/cm^3 and fat = 0.96 g/cm^3) and summed

(Martin, et al., 1989; Mungiole and Martin, 1990). Once the mass of each slice along the

leg was known, the center-of-mass, moment of inertia and radius of gyration were

calculated (Winter, 1990). Measurement validity was assessed using one intact bovine

shank containing fat, muscle, and bone. The MRI-derived bovine shank mass was 3.7%

(0.12 kg) less than the measured mass. The MRI-derived center of mass of the bovine

shank was computed to be 1.5% of the segment length (5.5 mm) more distal than the

actual center of mass.

MRI Hip Joint Center Location

In addition to the axial slices, coronal slices of the pelvis were imaged at 4 mm

intervals. Two slices were identified and exported into the image editing software for

measurement: the slice through the centers of the femoral heads, and the most anterior

slice in which the left anterior superior iliac spine (ASIS) was visible. Pelvic width was

calculated as the distance between the left and right ASIS. The displacement from the

left ASIS to the left hip joint center was then expressed in each anatomical direction as a percentage of the calculated pelvic width for use with our dynamic model, described below (21.5 ± 2.9% Posterior, 14.5 ± 2.6% Medial, & 32.8 ± 4.9% Inferior; mean ± SD)(Bauer, et al., 2005).

Tasks

Each subject performed no more than 20 drop landings from 61 cm and 20 repetitions of running approximately 20 m across the laboratory. For the drop landings, subjects were instructed to step off of a 61 cm tall platform, landing with one foot on each of two 40 x 60 cm force platforms (Model 4060-10, Bertec Corp., Columbus, OH 43229). The first 10 trials in which the subject successfully contacted the force platforms were used in the analysis. For the running trials, subjects were instructed to run at a self-selected pace, maintaining eye contact with a target positioned at head level at the opposite end of the lab. To prevent targeting effects, subjects were not made aware that they were to step on the force platform. The average running speed, determined using the position of the sacrum marker over time was 3.1 ± 0.2 m/s. The first 10 trials in which the subject successfully contacted the force platform with their left foot at the midpoint of the trial were used in the analysis. The task order was randomized. A 6-camera motion capture system (Mcam2, System 612, Vicon Peak, Lake Forest, CA, USA) sampled the positions of 20 lower extremity reflective markers at 120 Hz. Ground reaction forces were sampled from the left foot of each subject at 1080 Hz and synchronized with the motion capture data.

Joint angles and moments were calculated in 3D using a custom *BodyBuilder* program (Vicon Peak) and output at 540 Hz. Joint center locations and body segment orientations were computed from the marker data using transformations derived from anthropometric measurements and an initial trial of quiet standing. The 3D locations of the hip joint centers were determined for each subject using the MRI data in conjunction with the ASIS markers. Before processing, all marker data were filtered with a 4th order Butterworth no-lag filter at a cutoff frequency of 10 Hz, as determined by residual analysis (Winter, 1990).

Each trial was processed twice, once using body segment parameters estimated from the pediatric regression equations developed by Jensen (1986) and once using subject-specific thigh and shank parameters derived from the MRI data. Joint rotations, moments, and powers were determined about the mediolateral axis of the proximal segment, with joint power computed as the product of the joint moment and joint angular velocity. Key extrema in the moment and power sagittal-plane time histories during the stance phase of running (figure 2) and during the initial 200 ms contact phase of drop landings (figure 3) were averaged across trials for each subject. Moments and powers were normalized to body height (BH) and body weight (BW).

Statistics

Paired t-tests were used to assess differences in joint kinetic variables between methods within each activity. Differences were considered statistically significance if the p-value was less than 0.01.

3.4 Results

Running

Differences in body segment parameters resulted in statistically significant differences in sagittal-plane joint moments and powers during running. These differences were found in all of the moment extrema and in 3 of the 4 power extrema (Tables 1 & 2). However, the largest differences were small (0.3 ± 0.2 %BW*BH and 1.4 ± 0.8 %BW*BH/s for moments and power, respectively).

Drop Landing

As with running, differences in body segment parameters resulted in statistically significant differences in sagittal-plane joint moments and powers during drop landing from 61 cm. Significant differences were found in 3 of the 5 moment extrema and in 3 of the 5 power extrema (Tables 3 & 4). The largest difference in moments was 0.6 ± 0.3 %BW*BH. The largest difference in power was 4.2 ± 3.1 %BW*BH/s.

3.5 Discussion

From the premise that we previously found statistically significant differences between MRI and regression based body segment parameters (Bauer et al., in review) we sought to determine whether the differences in body segment parameters would significantly affect joint moments and powers during activities such as running and drop landings. We found that the differences in body segment parameters resulted in significant differences in joint moments and powers during both running and drop

landing. Nevertheless, from a practical standpoint, the differences in joint kinetics between methods are small.

Our findings are similar to a study investigating normal gait in children, where differences between adult cadaver-based body segment parameters and parameters derived from DXA in 7-13 year-old children (Ganley and Powers, 2004a) did not result in significant root-mean-square (RMS) errors between the joint moment curves. The differences in body segment parameters had a greater effect on joint moments calculated during the swing phase, where ground reaction forces did not enter into the joint kinetic calculations (Ganley and Powers, 2004b). However, in adults, variations in moments calculated using body segment parameters from six different studies have been reported to be over 19% at the hip during the stance phase of normal gait (Rao, et al., 2005). This large difference, compared to our smaller differences, is most likely due to the inclusion of six sources of body segment parameters, whereas there are few sources for pediatric body segment parameters for use in such a comparison. There are conflicting results in the literature concerning the effects of varying body segment parameters on joint kinetics calculated from tasks other than normal gait. In an adult population, RMS differences in hip flexion/extension moments of up to 6.7% and 11.3 % have been reported for stair ascending and descending, respectively, when comparing among five body segment parameter estimation methods (Fantozzi, et al., 2005). Additionally, one study found conflicting results while investigating body segment parameter variations on joint kinetics calculated from vertical jumps (Ho, et al., 2005). Lower extremity joint moments calculated using subject-specific body segment parameters were compared to moments calculated using body segment parameters based on cadavers, MRI, and gamma ray

scanning (Chandler, et al., 1975; Ho, et al., 2004; Zatsiorsky and Seluyanov, 1983). The authors found that the standard error of the estimate of ___ at the hip was similar between MRI and gamma ray prediction methods whereas the cadaver-based method had standard errors of the estimate that were up to 16 times greater than MRI or gamma ray. In contrast, researchers found no significant differences in the major knee joint force components calculated using adult male and female body segment parameters (de Leva, 1996) in 9-10 year old girls performing one-footed drop landings, (Sabick, et al., 2005). Thus, while it is clear that variations in body segment parameters do affect joint kinetics calculations, no studies have specifically addressed joint kinetics in activities other than normal gait when using body segment parameters developed specifically for use in young children.

Strengths and Limitations

To our knowledge, this study provides the first attempt to validate Jensen's (1986) age-based regression equations in estimating pediatric body segment parameters for use in joint kinetic calculations in girls performing activities other than walking. Because we processed a large number of trials for each activity and each subject, our data are likely representative of each subject's normal running and landing performance. To obtain a more precise center-of mass location and radius of gyration we used thinner slices (8.0 mm) compared to other studies (Ganley & Powers (2004) ~ 39 mm, Ackland et al. (1988) ~ 20 mm, Jensen (1986) ~ 20 mm, Durkin (2002) ~ 36 mm). In addition, we minimized the error in joint kinetics by measuring each subject's hip joint center location using MRI and applying the measurement to the rigid body model used in the joint kinetic

calculations. Of consideration, however, is that the body segment parameters measured using MRI in girls were compared to parameters derived from boys. The extent to which sex-differences may have contributed to the differences in parameters between estimation methods is unknown. Furthermore, our subjects were all very similar in age and body type. Thus, the applicability of our findings is limited to a narrow range of ages. Finally, no segment parameters were derived for foot the in this study. Since errors in joint kinetics tend to increase from distal to proximal joints, the differences we report may underestimate the differences we would have seen had our foot segment parameters been derived from MRI. However, the contribution of foot mass and inertial parameters to the kinetics of more proximal joints has been reported to be minimal (Krabbe, et al., 1997).

Implications

The advantage of using age-based regression equations to predict pediatric body segment parameters is that they are simple for everyone to use and require no special equipment. The body segment parameters estimated using age-based regression equations in this study appear to be a suitable substitute for more detailed anatomical imaging of 8-10 year old girls when quantifying joint moments and powers from running and drop landings. Thus, researchers investigating the mechanics of children participating in a wide variety of activities can be confident when using Jensen's regression equations to estimate pediatric body segment parameters.

3.6 Acknowledgements

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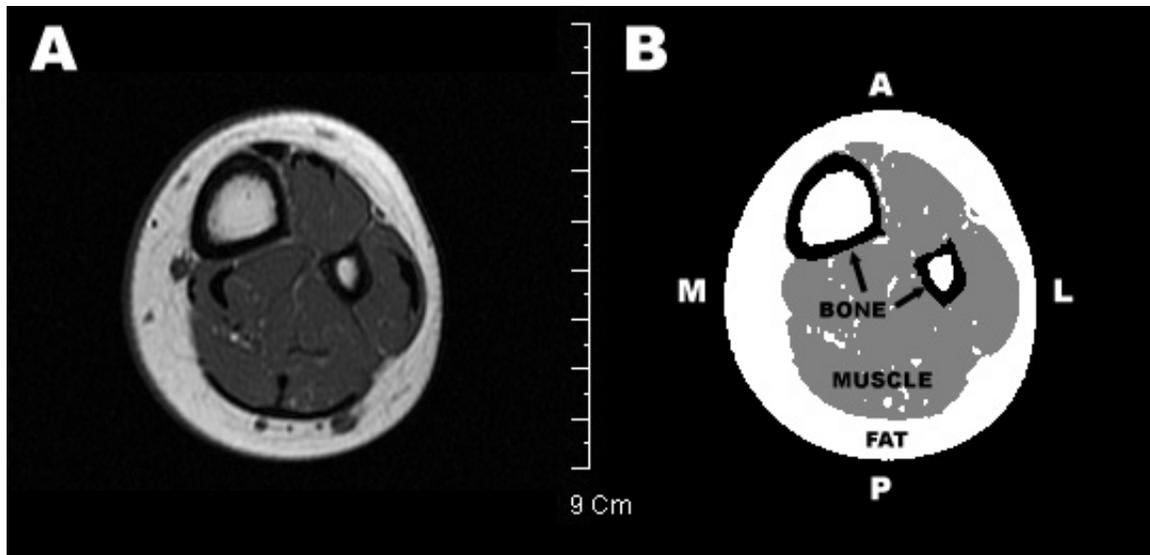
Figure 3.1. Inferior view of a representative slice through the shank. A) Original MRI, B) Processed slice depicting anatomical orientation and tissue types.
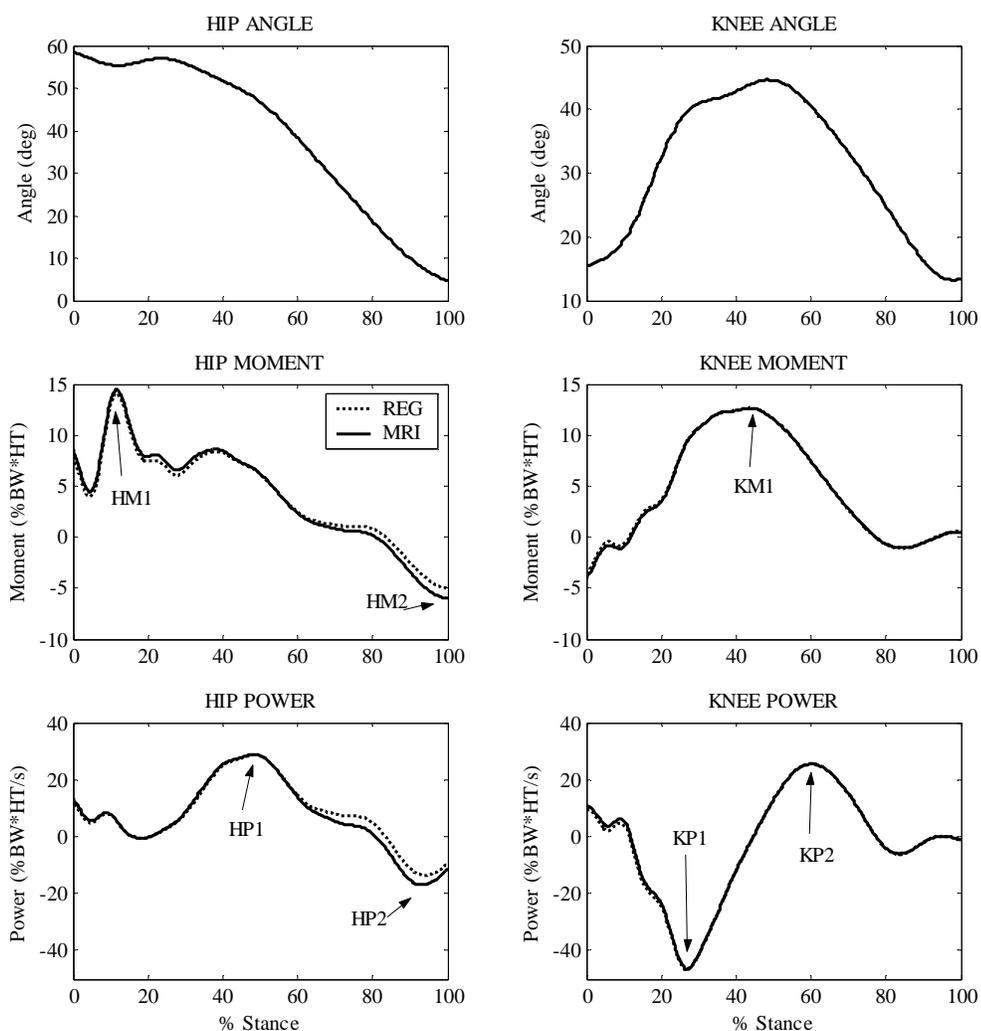
Figure 3.2. Representative running data showing extrema included in the analysis. Moments include: peak hip extensor moment (HM1) peak hip flexor moment (HM2), and peak knee flexor moment (KM1). Powers include: peak hip concentric power (HP1), peak hip eccentric power (HP2), peak knee eccentric power (KP1) and peak knee concentric power (KP2). NOTE: Ankle data are not presented as no body segment parameters were determined for the foot using MRI.
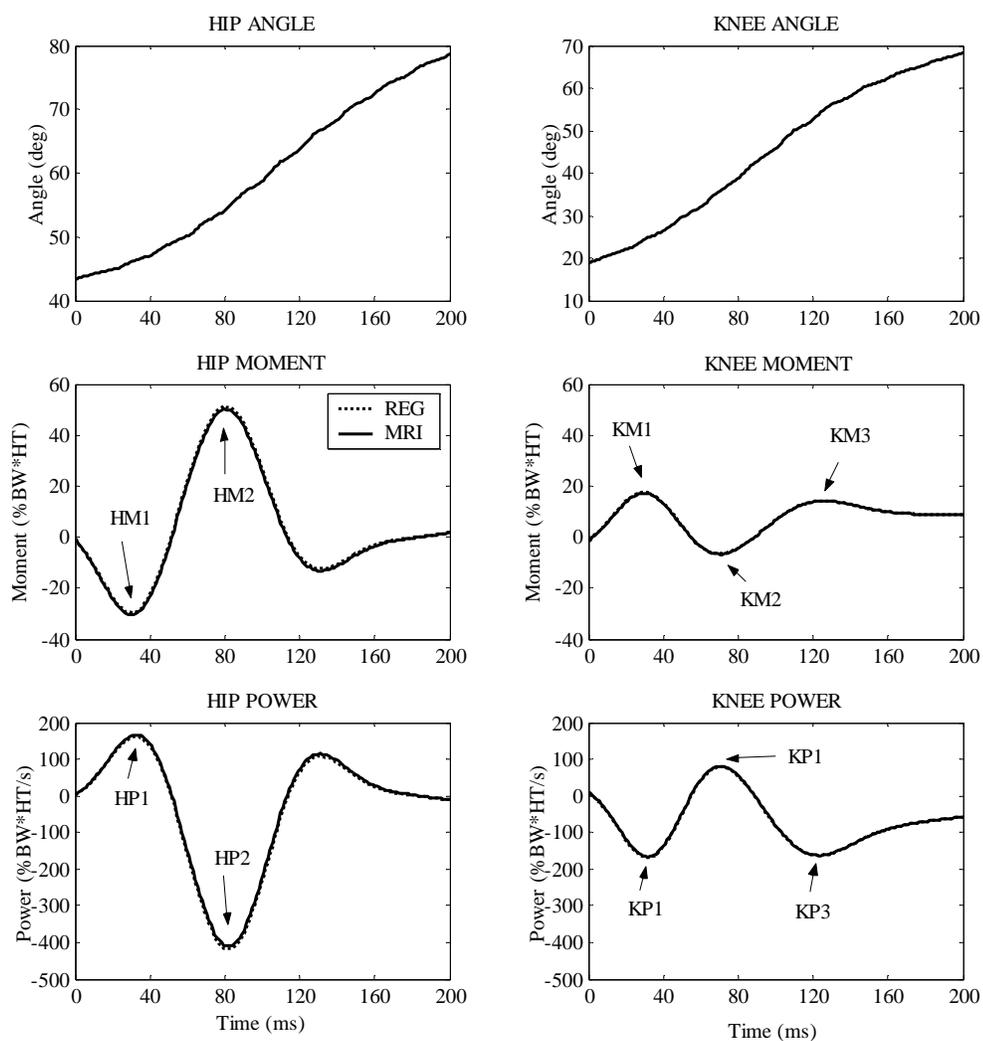
Figure 3.3. Representative drop landing data showing extrema included in the analysis. Moments include: peak hip extensor moment (HM1) peak hip flexor moment (HM2), peak knee flexor moment (KM1), peak knee flexor moment (KM2) and peak knee flexor moment (KM3). Powers include: peak hip concentric power (HP1), peak hip eccentric power (HP2), peak knee eccentric power (KP1), peak knee concentric power (KP2) and peak knee concentric power (KP3). NOTE: Ankle data are not presented as no body segment parameters were determined for the foot using MRI.

Table 3.1. Joint moment comparison between methods (RUN).

MOMENT (%BW*HT)	MRI	REGRESSION	P-VALUE
Peak Hip Extensor Moment (HM1)	20.95 ± 2.24	20.66 ± 2.30	0.000
Peak Hip Flexor Moment (HM2)	-4.00 ± 1.54	-3.73 ± 1.44	0.004
Peak Knee Flexor Moment 2 (KM1)	12.36 ± 2.08	12.32 ± 2.07	0.009

Table 3.2. Joint power comparison between methods (RUN).

| POWER (%BW*HT/s)	MRI	REGRESSION	P-VALUE
Peak Hip Concentric Power 1 (HP1)	32.99 ± 9.10	32.63 ± 8.97	0.005
Peak Hip Eccentric Power (HP2)	-13.45 ± 4.28	-12.07 ± 3.96	0.000
Peak Knee Eccentric Power 1 (KP1)	-60.01 ± 17.40	-60.12 ± 17.16	0.374
Peak Knee Eccentric Power 2 (KP2)	-34.30 ± 9.51	-34.04 ± 9.32	0.006

Table 3.3. Joint moment comparison between methods (DROP).

MOMENT (%BW*HT)	MRI	REGRESSION	P-VALUE
Peak Hip Extensor Moment (HM1)	-39.15 ± 11.11	-38.76 ± 11.13	0.000
Peak Hip Flexor Moment (HM2)	55.88 ± 16.75	56.46 ± 16.73	0.000
Peak Knee Flexor Moment 1 (KM1)	26.05 ± 6.91	26.17 ± 6.93	0.000
Peak Knee Extensor Moment (KM2)	-8.55 ± 3.88	-8.51 ± 3.86	0.015
Peak Knee Flexor Moment 2 (KM3)	16.71 ± 2.56	16.69 ± 2.56	0.161

Table 3.4. Joint power comparison between methods (DROP).

POWER (%BW*HT/s)	MRI	REGRESSION	P-VALUE
Peak Hip Concentric Power 1 (HP1)	226.80 ± 55.69	224.47 ± 55.95	0.000
Peak Hip Eccentric Power (HP2)	-492.52 ± 181.01	-496.69 ± 181.53	0.002
Peak Knee Concentric Power 2 (KP1)	-287.38 ± 66.76	-288.67 ± 66.79	0.000
Peak Knee Concentric Power 2 (KP2)	118.01 ± 56.98	117.46 ± 56.90	0.011
Peak Knee Eccentric Power 2 (KP3)	-219.24 ± 38.25	-218.87 ± 38.28	0.072

THE INTENSITY OF BONE LOADING AT THE HIP CAN BE DESCRIBED USING GROUND REACTION FORCES

Jeremy J. Bauer, Michael J. Pavol, Christine M. Snow and Wilson C. Hayes

4.1 Abstract

While exercise can relaibly be prescribed for improving heart and muscle health, there is no prescription for improving bone health in humans due to the absence of an intensity metric. Strain rate has been identified as a key intensity metric related to bone changes in animals (Turner, 1998). Thus, estimating loading rates from participating in several types of activities should address this lack of intensity metric in humans. We asked: How do hip joint reaction forces (HJRFs) and loading rates compare to ground reaction forces (GRFs) and loading rates within activities? How do forces, loading rates and moments compare among activities? Ten girls (9.6 ± 0.9 yrs, 37.0 ± 7.1 kg, 141.0 ± 9.4 cm, mean ± SD) performed 10 trials of walking, running and drop landing from 30, 46 & 61cm across two forceplates. Maximum GRFs, HJRFs and their respective loading rates were quantified for each activity. All ground and hip variables were linearly related. GRFs from walking were less than other activities. GRFs from 61 cm drop landings were greater than other activities. Each of the drop landings were different from each other. Loading rates at the ground during walking were less than other activities. Loading rates during running were greater than walking and less than drop landings. There were no differences in loading rates among drop landings. Extensor moments at the hip were significantly different among all conditions, except between drop landings from 30 and 46 cm. Flexor moments from drop landings were significantly different among all conditions. By sampling ground reaction force data and calculating the loading rate at the ground, we have presented an easy way for researchers to quantify their regimen of activities in terms of bone loading, allowing for a more detailed relationship between bone loading and changes in bone mass to be developed in humans.

4.2 Introduction

Osteoporosis is a disease of crisis proportions in the United States, with more than 700,000 vertebral and 300,000 hip fractures annually (National Osteoporosis Foundation, 2002). Approximately 7.8 million women and 2.3 million men in the U.S. have osteoporosis and another 21.8 million women and 11.8 million men have low bone mass. These numbers are expected to double by 2050. Based on studies in children where gains in bone mass achieved over a 7-month training program were maintained after 7 months of detraining (Fuchs and Snow, 2002), the National Institutes of Health (NIH) has emphasized that increasing peak bone mass during youth is essential to preventing fractures later in life. However, there is currently no clear prescription for increasing peak bone mass in youth

A majority of human based research in bone and exercise has used broad programs of multiple resistance training exercises and cardiovascular regimens to increase bone mass (Heinonen, et al., 1998; Kerr, et al., 2001; Kerr, et al., 1996; Kohrt, et al., 1997; Kohrt, et al., 1995; Lohman, et al., 1995; Nelson, et al., 1994; Pruitt, et al., 1992). However, these broad programs do not allow clinicians to develop a relationship between the loading stimulus and changes in bone mass since the intensity component of the exercise does not directly relate to bone loading. Additionally, since multiple forms of loading are used, it is nearly impossible to associate one type of loading with changes in bone. The justification for the mode of exercise used has been based on muscle and cardiovascular based programs. For example, the intensity metric for the cardiovascular system and skeletal muscle is a percentage of maximum heart rate or maximum VO_2 and a percentage of 1 repetition maximum (1 RM) or 6-8 RM, respectively. The intensity

metric for bone in humans has yet to be defined. Fortunately, animal studies have provided a wealth of information through randomized controlled trials concerning variables associated with osteogenesis.

Using an animal model, strain rate (rate of bone deformation) has been identified as the key dependent variable related to the osteogenic response of bone (Turner, 1998). Strain rate is a function of the magnitude of strain (bone deformation) and the frequency of loading. However, strain is measured directly on the bone in animals, and cannot be ethically measured on the human femur. Therefore, a metric based on external forces that cause bone to deform and their rate of application has been proposed previously, but not has not been widely used (Bauer, et al., 2001; Turner and Robling, 2003). Thus, in order to better derive exercise regimens aimed at optimizing hip bone mass and preventing fractures, it is important to be able to relate a measurable stimulus (force or loading rate) to a bone response.

The purpose of this research is to describe and compare several common childhood activities in terms relating to bone loading. By estimating the loading at the hip using analogous variables to strain and strain rate such as hip joint reaction forces and loading rates at the hip, we hope to provide a means for bone researchers to begin better defining the relationship between exercise and changes in bone mass. Importantly, in order to make the hip loading estimates easy for clinicians to derive, we wish to know whether ground reaction forces can be used to describe the loading at the hip. Specifically, we wish to answer: 1) How do the resultant reaction forces at the hip compare to ground reaction forces within walking, running and drop landings from 30 cm, 46 cm and 61 cm? 2) How do loading rates at the hip joint compare to loading rates

at the ground within each task? 3) If there is a significant linear relationship between ground reaction forces and hip joint reaction forces, how do the ground reaction forces and loading rates at the ground compare among activities? And 4) based on recent studies relating joint moments to local bone mass measures, how do sagittal plane moments at the hip compare among activities (Hurwitz, et al., 1998a; Hurwitz, et al., 1998b)?

4.3 Methods

Subjects

Ten girls (age: 9.6 ± 0.9 yrs, mass: 37.0 ± 7.1 kg, height: 141.0 ± 9.4 cm, BMI: 18.5 ± 1.8; mean $\pm$ SD) were recruited. Each subject and her parent or guardian provided written informed consent to participate. Institutional Review Board approval was obtained.

Data Collection/Processing

Each subject performed no more than 20 repetitions of the following activities: walking and running approximately 20 m across the laboratory and drop landing from 30 cm, 46 cm and 61 cm. For the walking and running trials, subjects were instructed to proceed at a self-selected pace, maintaining eye contact with a target positioned at head level at the opposite end of the lab. To prevent targeting effects, subjects were not made aware that they were to step on the force platform. The first 10 trials in which the subject successfully contacted a force platform with their left foot at the midpoint of the trial were used in the analysis.   The average speed of progression, determined using the position of the sacrum marker over time was 1.5 ± 0.1 m/s and 3.1 ± 0.2 m/s for walking

and running, respectively. For the drop landings, subjects were instructed to step off a 30

cm, 46 cm and 61 cm platform, landing with one foot on each of two 40 x 60 cm force

platforms (Model 4060-10, Bertec Corp., Columbus, OH 43229). The first 10 trials at

each height in which the subject successfully contacted the force platforms were used in

the analysis. The order of performing the tasks was randomized. A 6-camera motion

capture system (Mcam2, System 612, Vicon Peak, Lake Forest, CA, USA) sampled the

positions of 20 lower extremity markers at 120 Hz. Ground reaction forces were sampled

from the left foot of each subject at 1080 Hz and synchronized with the motion capture

data.

A rigid body model was used to represent the three segments of each subject's

lower extremity (Figure 4.1). A custom *BodyBuilder* program (Vicon Peak) was written

to calculate 3D joint reaction forces and moments from the ground reaction forces and

accelerations of the segments using inverse dynamics. The data were output at 540 Hz.

Joint center locations and body segment orientations were computed from the marker

data using transformations derived from anthropometric measurements and an initial trial

of quiet standing. Subject-specific segment mass, center-of-mass location and inertia of

the segments required for the joint reaction force and moment calculations, were derived

from MRI and have been reported previously (Bauer et al., *in review*). The 3D locations

of the hip joint centers were determined for each subject using MRI data in conjunction

with the external ASIS markers. Before processing, all marker data were filtered with a

4th order Butterworth no-lag filter at a cutoff frequency of 10 Hz, as determined by

residual analysis (Winter, 1990).

The force traces from each task contain two force peaks: P1 and P2 (Figure 4.2). During walking and running, P1 is representative of heel strike and generally described as the "impact" peak whereas P2 is commonly termed the "active" peak which occurs during propulsion. During drop landings, P1 is due to initial toe contact whereas P2 occurs near heel strike. Four variables were quantified at P1 and P2: 1) maxGRF, 2) GRF loading rate, 3) maxHJRF, and 4) HJRF loading rate. Loading rate was quantified by calculating the slope, from 20-80%, of the rising portion of each peak (shaded portion of Figure 4.2). Additionally, for each activity, only the maximum value of P1 and P2 was used in comparing among activities. Finally, we quantified peak hip flexion and extension moments in the sagittal plane.

Statistics

Linear regression was to assess the relationship between ground reaction forces and hip joint reaction forces as well as between loading rates at the ground and hip. A within-subjects repeated measures ANOVA was used to determine whether differences in the maximum forces, loading rates and moments exist among the 5 activities. The Bonferroni method was used post-hoc to determine where the differences, if any, occurred. SPSS version 13.0 was used to perform all statistical calculations (SPSS, Inc., Chicago, Illinois 60606). Statistical significance was set at $p < 0.05$.

4.4 Results

Using P1 and P2 from each activity, the relationship between maxGRF and maxHJRF was linear (HJRF = 0.821xGRF – 0.037; $R^2 = 0.988$, $p < 0.001$; Table 4.1;

Figure 4.3). The relationship between maximum loading rates at the ground and hip was also linear (HJRFrate = 1.090xGRFrate – 4.977; R^2 = 0.981, p < 0.001; Figure 4.4). Therefore, given the significant linear relationship between ground and hip joint variables, only the ground based variables were used to compare forces and loading rates among activities.

Ground Reaction Forces

Ground reaction forces from walking were significantly lower compared to the other activities, whereas forces from 61 cm drop landings were significantly greater than all other activities (Figure 4.5). Each of the drop landings were different from each other, increasing in force as the drop height increased. However, maximum forces from running were not significantly different than drop landings from 30 cm and 46 cm.

Ground Loading Rates

Loading rates at the ground during walking were significantly lower compared to all other activities (Figure 4.6). Loading rates during running were significantly greater than walking and less than each of the drop landing conditions. There were no significant differences in loading rates among the three drop landing conditions.

Hip Moments

Extensor moments at the hip were significantly different among all conditions, with the exception of one non-significant comparison between drop landings from 30 cm and 46 cm (Figure 4.7). Flexor moments from drop landings were significantly different

among all conditions. However, flexor moments were similar between walking and running.

4.5 Discussion

The purpose of this research was to describe and compare several common childhood activities using variables specifically related to skeletal loading. Additionally, we sought to determine whether or not indirectly inferred calculations of hip joint loading could be explained using more directly measured forces at the ground. We found significant linear relationships between the ground and hip variables, thus making it unnecessary to calculate hip joint reaction forces when describing resultant loading at the hip. While the forces calculated at the hip in our study do not consider the contribution of muscle activity, thus underestimating the real joint force, data from two studies have shown that *in vivo* hip forces are consistently greater than ground reaction forces (Bassey, et al., 1997; Bergmann, 2001). In one study, a patient with an instrumented hip replacement performed jogging, slow jumping and fast jumping on a force platform with and without shoes (Bassey, et al., 1997). The peak forces at the hip were 1.7 ± 0.5 BW greater than those measured at the ground. More importantly, and similar to our findings, the loading rates at the hip were not significantly different than those at the ground. In another study, four subjects with instrumented hip replacements walked at a slow, normal and fast pace over a force plate (Bergmann, 2001). The peak forces at the hip were 1.3 ± 0.1 BW greater than those measured at the ground and the loading rates at the hip, like the previous study and our current study, were not significantly different from those at the ground. Thus, even in the ideal case of having *in vivo* joint forces, ground reaction

forces and loading rates at the ground can be used to describe the resultant loads and loading rates at the hip.

Another purpose of this study was to quantify several activities in terms of a bone loading metric that has previously been proposed for comparison with changes in bone mass (Bauer, et al., 2001). Two of the activities were normal daily activities: walking and running. One activity, drop landings from 61 cm, has previously been associated with up to 4.5% changes in bone mineral content in children (Fuchs, et al., 2001). The remaining two activities, drop landings from 30 and 46 cm, were included based on pilot data showing that a 50% reduction in drop height decreased the maximum force, but had no effect on loading rate (Bauer and Snow, 2001). We found that drop landings from 61 cm had significantly greater forces and loading rates compared to more typical activities such as walking and running. Theoretically, the higher forces and loading rates would make drop landings a better candidate for causing osteogenesis compared to walking or running. Interestingly, since strain rate is the variable most related to changes in bone mass in animal models, and externally measured loading rates are analogous to strain rate, then drop landings from any of the investigated heights should be similarly osteogenic. Of particular interest is that the forces from 30 and 46 cm drop landings were similar to running, whereas the loading rates from 30 and 46 cm landings were similar to those from 61 cm drop landings, which have been related to osteogenesis in children. Thus, perhaps an exercise regimen of drop landings from heights above 30 cm included in an exercise program would be enough to stimulate bone osteogenesis. It should be noted that numbers of repetitions and rest between loading sessions must also be considered in designing an exercise protocol for bone.

The final purpose of this research was to emphasize the need for relating exercises to changes in bone mass so that it is easier to prescribe exercise specific to increasing bone mass. As discussed previously, most research concerning exercise and its effects on bone mineral accrual in children has involved broad programs of multiple resistance training exercises and cardiovascular regimens. For example, thirty-eight 9½ year old children increased BMD at the hip and spine by participating in a ten month intervention including the following activities: Twenty-station weight circuit, "high-impact" aerobics, modified soccer, skipping, ball games, dance, modified soccer and more (Morris, et al., 1997). Note that the term "high-impact" is subjective and does not adequately describe skeletal loading. In another study, twenty 10 year old children increased BMD at the hip and spine by participating in an eight month intervention including the following activities: Basketball, weight training, aerobics, soccer, volleyball, gymnastics and folk and line dancing (Bradney, et al., 1998). Human based studies are difficult to execute due to the complexities of human physiology, social requirements such as limited time and funding and the need for activities to be beneficial not only to bone, but muscle and cardiovascular health as well. While the studies mentioned above certainly contribute to the knowledge base for exercise effects on bone, the intensity as it relates to bone was not addressed, thus taking us no closer to developing a relationship between loading and changes in bone mass in humans. Before we can prescribe individualized exercise programs for bone health, we need to observe the bone response resultant from exposure to only one mode of mechanical loading or exercise. This has been performed in one pediatric population, but needs to be performed with other types of loading and in different populations (Fuchs, et al., 2001).

Strengths and Limitations

Because we processed a large number of trials for each subject and activity, our data were likely representative of each subject's normal performance. We were also able to minimize error in joint kinetics by applying subject-specific body segment parameters and hip joint center locations to the rigid body model. However, only girls were tested. The extent to which the observed differences between activities were specifically related to sex is therefore unknown. Our subjects were also very similar in age and body type. Thus, our results can only be applied to a small pediatric population. Finally, the hip joint reaction forces and loading rates used in the regression equations do not take into consideration the muscle forces acting across the hip joint. Thus, the hip joint reaction forces do not completely represent the loading environment at the hip. However, until we can directly measure strain on the human femur *in vivo*, externally measured loading rates are the closest approximation available.

Implications

Human studies examining the relationship between exercise and changes in bone mass have been lacking one fundamental component, a description of the mechanical loading stimulus, or intensity as it relates to bone. One of the first mentions of the need to quantify the intensity of loading as it relates to changes in bone mass came from the development of the Daily Stress Stimulus (Carter, et al., 1987). The Daily Stress Stimulus was developed to quantify the mechanical stimulus in the bone by combining the influence of the number and magnitude of individual loading cycles over the course

of a day. However, studies testing the stress stimulus using animals have found that strain rate is a better predictor of changes in bone compared to the strain magnitude (Adams, et al., 1997; Qin, et al., 1998). Additionally, the difficulty with applying the Daily Stress Stimulus to humans is that there is still a need to estimate the stresses imparted at particular skeletal sites such as the hip, which cannot ethically be measured in humans. Taking a similar approach, one group proposed a means through which to derive an exercise prescription based on an Osteogenic Index (OI)(Turner and Robling, 2003). The OI was developed based on the knowledge gained through animal studies where the relationship between strain rate, the number of applied loads, and rest cycles and changes in bone was determined. However, while the authors account for every component of an exercise prescription and even state that the intensity may be estimated as the peak ground reaction force divided by the duration of the ground reaction force, or the loading rate, they end up using only the peak ground reaction force as the intensity variable in their examples, thereby ignoring their own research demonstrating that strain rate is the most important loading variable related to changes in bone (Turner, 1998). If only peak ground reaction forces were used to calculate the OI, according to our study, running and drop landing from 30 cm and 46 cm would be similarly osteogenic. However, the OI for drop landings would be over twice that of running for the same number of loading cycles and rest cycles if loading rate was used as the intensity variable. Thus, by merely sampling ground reaction force data and calculating the loading rate at the ground have presented an easy way for bone/exercise researchers to quantify their regimen of activities in terms of bone loading, allowing for a more detailed relationship between bone loading and changes in bone mass to be developed in humans.

4.6 Acknowledgements

4.7 References

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FIGURES

F_{hip_y}

M_{hip}

THIGH

F_{hip_x}

W_t

$m_t\,a_{tx}$

$I_t\,\ddot{\theta}_t$

$m_t\,a_{ty}$

F_{knee_x}

M_{knee}

F_{knee_y}

F_{knee_y}

F_{knee_x}

M_{knee}

SHANK

W_s

$m_s\,a_{sx}$

$I_s\,\ddot{\theta}_s$

$m_s\,a_{sy}$

F_{ankle_y}

W_f

FOOT

F_{ankle_x}

F_{ankle_x}

$m_f\,a_{fx}$

M_{ankle}

M_{ankle}

$I_f\,\ddot{\theta}_f$

F_{ground_x}

F_{ankle_y}

M_{ground}

$m_f\,a_{fy}$

F_{ground_y}

F = reaction force
M = moment
W = weight
I = inertia
$\ddot{\theta}$ = angular acceleration
m = mass
a = acceleration
t = thigh
s = shank
f = foot
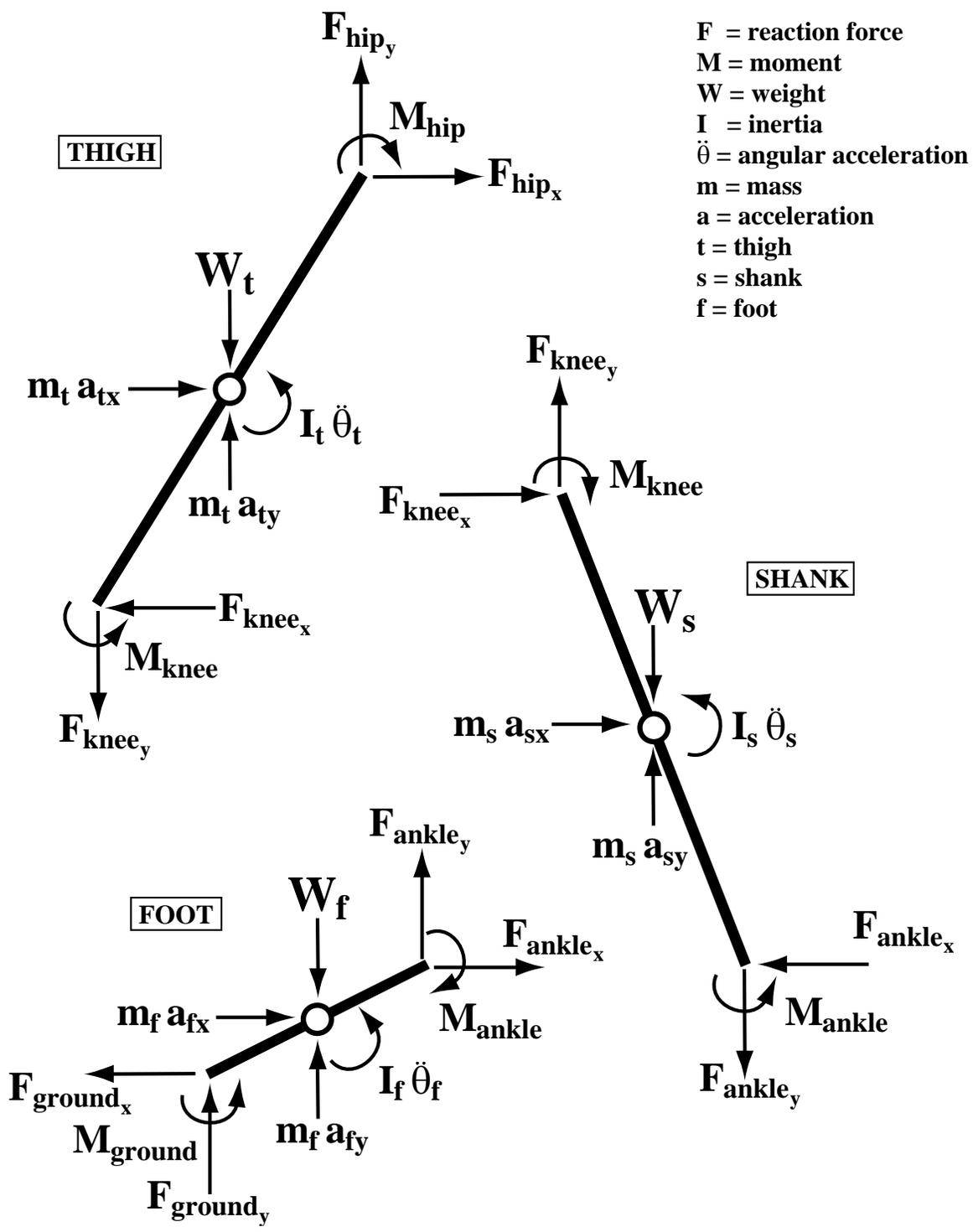
Figure 4.1. Rigid body model representing the lower extremity.
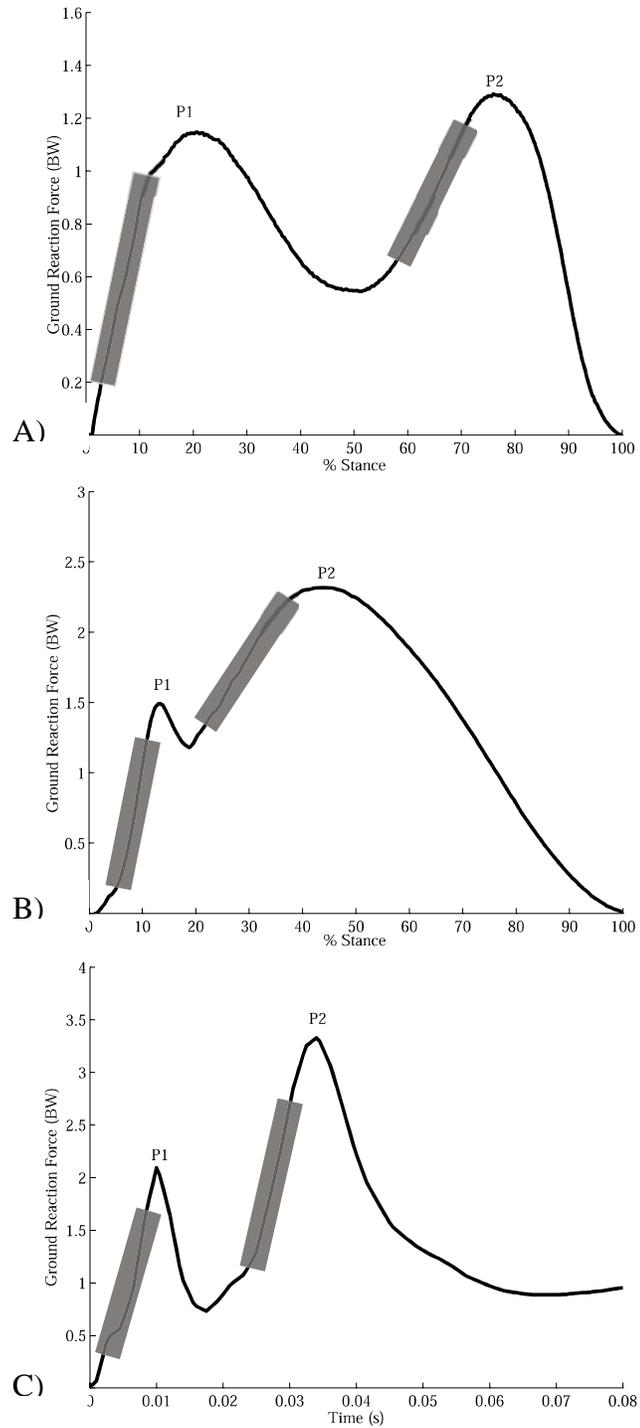
FIGURE 4.2. Representative ground reaction force trace from A) walking, B) running and C) drop landing from 46 cm. The shaded area represents the portion of the trace used in calculating loading rate.
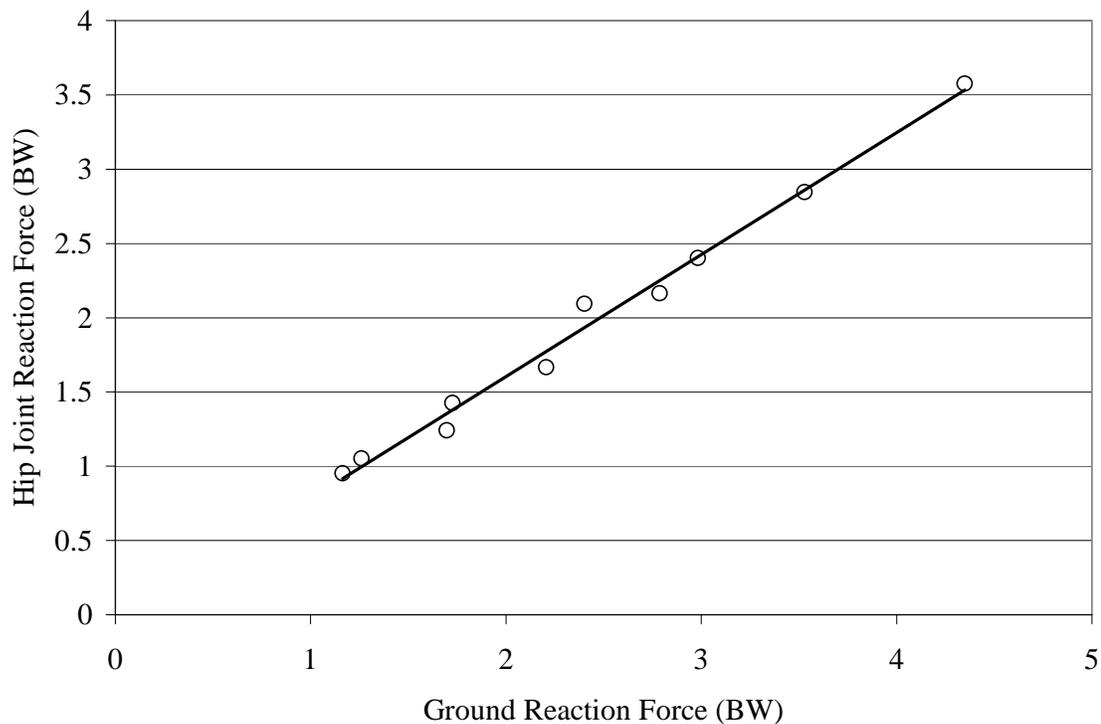
FIGURE 4.3. Relationship between resultant ground reaction forces and resultant hip joint reaction forces (HJRF = 0.821xGRF – 0.037; R^2 = 0.988).
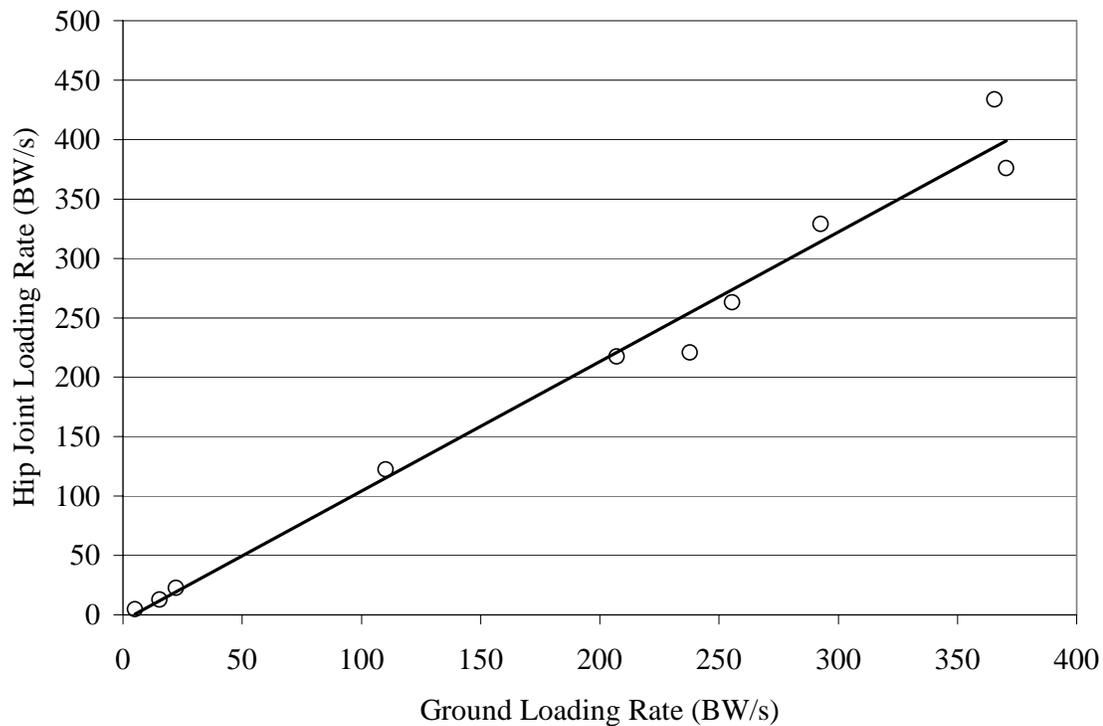
FIGURE 4.4. Relationship between measured loading rate at the ground and calculated loading rate at the hip (HJRFrate = 1.090xGRFrate – 4.977; R^2 = 0.981).
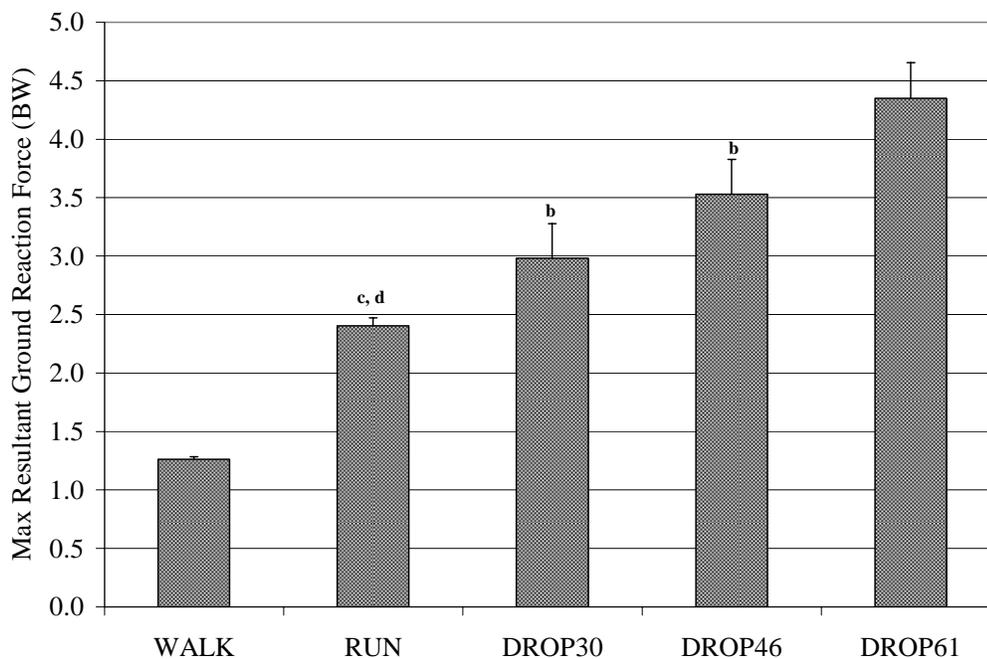
FIGURE 4.5. Resultant ground reaction forces compared among activities. (a = same as WALK, b =same as RUN, c = same as DROP30, d = same as DROP46 and e = same as DROP61; p< 0.05)
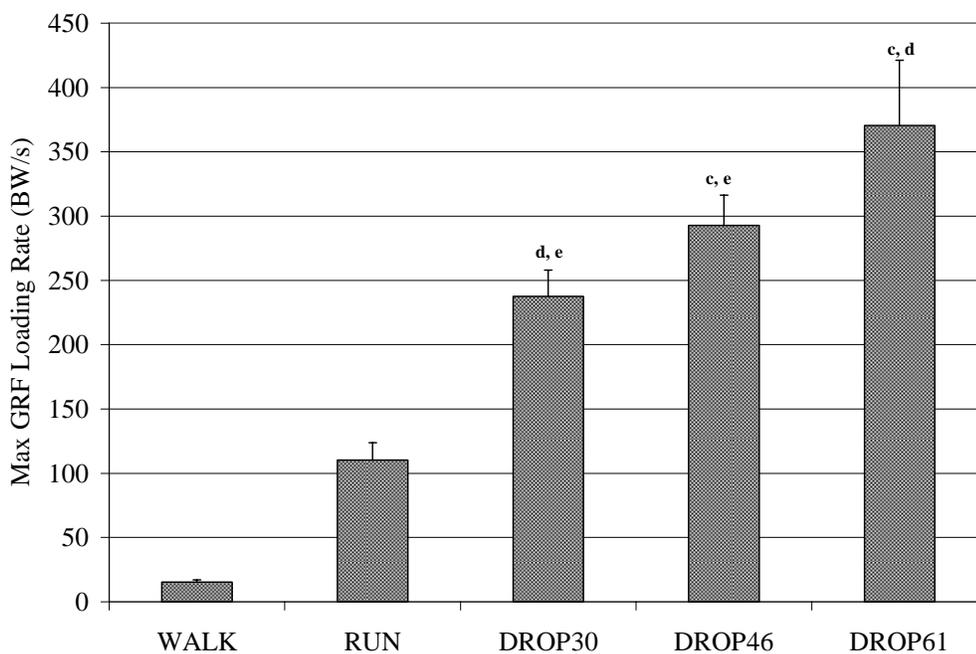
FIGURE 4.6. Loading rates at the ground compared among activities. (a = same as WALK, b =same as RUN, c = same as DROP30, d = same as DROP46 and e = same as DROP61; p< 0.05)
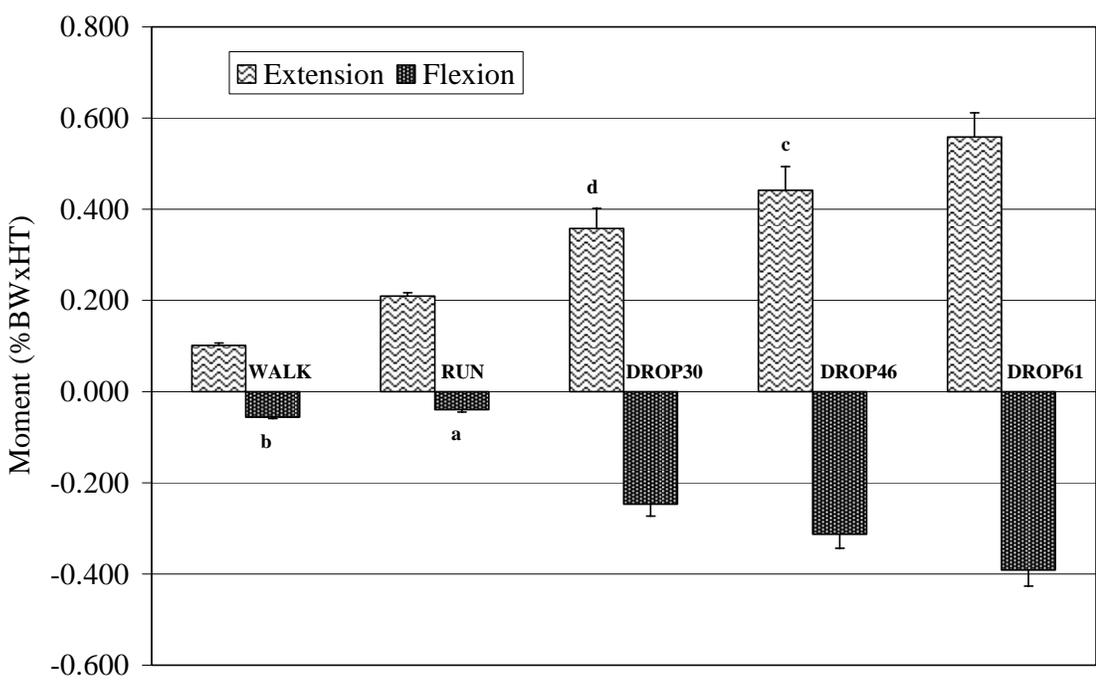
FIGURE 4.7. Sagittal plane hip joint moments compared among activities. (a = same as WALK, b =same as RUN, c = same as DROP30, d = same as DROP46 and e = same as DROP61; p< 0.05)

TABLES

TABLE 4.1. Summary force and loading rate data. Bolded value is the maximum of P1 and P2 for each activity and variable (MEAN ± SD).

| PEAK_ACTIVITY	GRF (BW)			HIP (BW)			GRFrate (BW/s)			HIPrate (BW/s)		
P1_WALK	**1.3**	**±**	**0.1**	**1.1**	**±**	**0.1**	**15.4**	**±**	**5.1**	**13.0**	**±**	**4.3**
P2_WALK	1.2	±	0.1	1.0	±	0.1	5.1	±	1.3	4.7	±	1.2
P1_RUN	1.7	±	0.3	1.4	±	0.2	**110.2**	**±**	**42.5**	**122.3**	**±**	**50.2**
P2_RUN	**2.4**	**±**	**0.2**	**2.1**	**±**	**0.2**	22.3	±	4.1	22.5	±	4.1
P1_DROP12	1.7	±	0.4	1.2	±	0.4	**237.7**	**±**	**64.5**	**220.6**	**±**	**89.5**
P2_DROP12	**3.0**	**±**	**0.9**	**2.4**	**±**	**0.9**	207.1	±	102.4	217.5	±	109.5
P1_DROP18	2.2	±	0.5	1.7	±	0.5	**292.6**	**±**	**74.4**	**329.0**	**±**	**131.7**
P2_DROP18	**3.5**	**±**	**0.9**	**2.8**	**±**	**0.9**	255.5	±	117.6	263.0	±	118.8
P1_DROP24	2.8	±	0.6	2.2	±	0.6	365.6	±	109.7	**433.8**	**±**	**182.6**
P2_DROP24	**4.3**	**±**	**1.0**	**3.6**	**±**	**0.9**	**370.5**	**±**	**160.6**	375.9	±	160.4

CONCLUSION

5.1 Summary

The goal of the research presented in this dissertation was three-fold: 1) to fill a void in the literature concerning the estimation of body segment parameters in children, 2) to identify an easy way for bone researchers and clinicians to describe the intensity of an exercise in terms specific to skeletal loading and 3) to emphasize the need for bone researchers to quantify their exercise programs in terms relating to skeletal loading in order to develop a detailed relationship between skeletal loading and changes in bone mass. Prior to the research presented in this dissertation, no studies had published a validated means of predicting segment parameters in children. By using MRI, a technique that has been shown to be valid in estimating body segment parameters, we were able to assess the validity of previously published methods for estimating segment parameters in children. We found that the greatest differences in joint moments and powers due to differences in body segment parameters were not practically significant. Thus, given the confidence that pediatric segment parameters could be easily estimated using regression equations, we sought to pursue an easily measurable intensity variable that clinicians and researchers could use to describe skeletal loading.

Researchers studying bone in animals have shown that dynamic bone deformation, specifically, the rate of bone deformation is significantly related to changes in bone mass. Thus, we sought to non-invasively quantify the forces and loading rates that cause bone deformation in humans. Specifically, in focusing on developing a simple method for clinicians and researchers to use in describing the intensity of skeletal loading, we sought to determine whether simply measured forces at the ground could describe joint reaction

forces at the hip. We found significant linear relationships between ground and hip variables. Thus, allowing for the estimation of hip loading using forces measured at the ground. Our findings are strengthened by results from previous studies where forces from instrumented hip prostheses were linearly related to forces at the ground. Once we determined that ground reaction force data could be used to describe the forces and loading rates at the hip, we sought to quantify several activities in order to assess their potential for increasing bone mass. We found that walking and drop landings from 61 cm had the lowest and highest forces, respectively and that all drop landings were different from each other, increasing in force as height increased. Additionally, we found that forces from running were not different than 30 and 46 cm landings. However, in the animal literature, the rate of loading is the crucial variable in determining whether there will be an osteogenic (bone building) response. We found that loading rates at the ground during walking were less than other activities and that loading rates during running were greater than walking and less than drop landings. However, there were no differences in loading rates among drop landings. The lack of differences in loading rates among drop landing conditions is important since drop landings from 61 cm have been shown to increase bone mass in children. Thus, if loading rate is indeed the variable most related to changes in bone mass, then drop landings ranging from 30 to 61 cm would be similarly osteogenic. In addition, drop landings would be easy to implement into any exercise program. Thus, by quantifying ground forces and loading rates, we have presented an easy way for researchers to quantify skeletal loading intensity, allowing for a more detailed relationship between loading and changes in bone mass to be developed.

The final goal of this research was to emphasize the need to be able to relate skeletal loads from various exercises to changes in bone mass so that it is easier to prescribe exercise specific to increasing bone mass. As discussed previously, a majority of research concerning exercise and its effects on bone mineral accrual in children has used broad programs of multiple resistance training exercises and cardiovascular regimens. While those studies do contribute to the knowledge base for exercise effects on bone, the intensity as it relates to bone was not addressed, thus taking the field of human bone research no closer to developing a relationship between loading and changes in bone mass. Before we can prescribe individualized exercise programs for bone health, we need to observe the bone response resultant from exposure to only one mode of skeletal loading or exercise. This has been performed in one pediatric population, but needs to be performed with other types of loading and in different populations.

5.2 Future Considerations

In a best case scenario researchers would be able to perform controlled bone loading studies in human bone where direct measures of bone strain (deformation) could be compared to a detailed bone response. However, measuring bone strain in humans at the hip and spine is invasive and unethical with the methods currently available. In addition, a measure of detailed structural bone response would require high dose X-ray sources such as those used in computed tomography (CT) scans. Delivering these high x-ray doses in normal populations not requiring the x-rays for treatment is also unethical. Therefore, future studies must continue to develop methods that will non-invasively provide some insight into the response of human bone to mechanical loading.

Predicting strains in the hip requires knowledge of the bony geometry, material properties and forces causing the strain. Bony geometry of the hip can be determined using CT scans, DEXA or MR imaging. While DXA only provides geometry in two-dimensions, MRI and CT provide 3D geometry. However, as noted earlier the high radiation dose of the CT prevents this method from being used in normal populations. MRI does not directly image bone, but bony geometry can be measured based on the soft tissue surrounding the bone. Therefore, it is possible to get a fairly detailed representation of the local geometry of the hip. Currently, the only way to know the true material properties of the proximal femur is to break cadaveric femurs in a controlled loading environment outside of the body. However, researchers investigating the mechanical properties of trabecular bone found that the strength and modulus of trabecular samples varied as the square and cube of bone apparent density respectively (Hayes et al. 1991). In addition, BMD of the proximal femur has been shown to explain 92% of the failure load at the same site (Courtney et al. 1995). Therefore, it may be feasible to merely estimate mechanical properties of the hip *in vivo* using DXA. However it is important to note that bone material properties vary according to the direction and rate of loading and that physiologic loading rates were not used in either study.

The forces causing deformation of the proximal femur are a combination of externally applied forces and internally generated forces (i.e. muscle contraction across the joint). Externally applied forces can easily be quantified with force platforms. Whereas internally applied forces prove to be the crux of the problem. Muscles are actuators within the body that cause motion in the limbs and consequently joint loading. The muscle contribution to loading within the shaft of the femur has been shown to range

from 200-300 N in a subject lying at rest (Bassey et al. 1997). In addition, numerous

studies have shown that heavy resistance training can increase BMD at both the hip and

spine. Therefore, it is clear that muscle forces contribute significantly to bone loading.

However, the problem lies in how to quantify the muscle force contribution. The

determination of forces acting at the hip joint due to muscle requires some knowledge of

muscle anatomy (i.e. origin, insertion, cross-sectional area, moment arm and angle of

pennation), muscle dynamics (i.e. activation timing, activation magnitude and

viscoelastic properties), and mechanical properties of other tissues surrounding the joint

such as articular cartilage, ligaments, tendons, and fat (i.e. heel pad contact properties).

Muscle origins and insertions have been measured in cadavers and average locations have

been scaled to various skeletal sizes in software designed to investigate the effects of

muscle length changes on human gait (Delp et al. 1990; Brand et al. 1982). Muscle

cross-sectional area can easily be measured using MRI. However, the moment arm, or

the perpendicular distance from a muscle crossing a joint to the joint center varies

depending on the position of the joints. Therefore, while moment arms can be quantified

using MRI, it is important to quantify the moment arms through the complete range of

motion of the limb as has been done previously for the finger and hip (Fowler et al. 2001;

Delp et al. 1999). Angle of pennation for each muscle can be determined using

ultrasonography (Narici et al. 2003). Concerning muscle dynamics, activation timing can

be determined to some extent by measuring EMG either on the surface of the skin over a

certain muscle or by needle electrodes inserted into the muscle. In addition, activation

level can be quantified by comparing an EMG signal measured during a maximum

voluntary contraction to a contraction measured during a task of interest. However, both

EMG methods provide noisy signals and poor quality data concerning deep musculature. Thus, the noise in the EMG signals prevents the direct comparison of individual muscle EMG data and prevents any valid measure of muscle force. However, EMG is very useful if the desire is to only know when a muscle is active or inactive. In contrast to EMG uncertainties, researchers are much more confident with the current knowledge in muscle mechanical properties for modeling purposes (Winters and Woo, 1990). Material properties of articular cartilage, ligaments, tendons and other tissues are better documented, but still under investigation (Jurvelin et al. 2003; Steiner et al. 1994; Noyes et al. 1984; Maganaris 2002; Miller-Young et al. 2002). Thus, only when every detail is known about the anatomy and physiology of the structures in and around the hip joint can an ideal model be created to predict strains in the proximal femur due to mechanical loading.

Predicting joint forces via computer models based on detailed measures of anatomy *in vitro* (cadavers), anatomy *in vivo* (MRI, CT, etc.), skeletal movement and external forces (force platforms) is the ideal next step to evaluating bone loading in humans non-invasively. However, unless joint forces are quantified *in vivo*, there is no way to validate the models. Fortunately, many researchers collaborating with surgeons have taken advantage of joint replacement procedures to investigate hip joint forces through means of instrumented prostheses (Bassey et al. 1997; Bergmann et al. 1988; Bergmann et al. 2001; Bergmann et al. 1993; Bergmann et al. 1997; Bergmann et al. 1995; Bergmann et al. 1984; Rydell 1966; English and Kilvington 1979; Davy et al. 1988; Taylor and Walker 2001; Pedersen et al. 1997). For years researchers have been developing 3D models to predict joint forces during human gait. These models use

muscles to move segments through the range of motion in an attempt to accurately represent human movement. The models are often validated based on muscle activation timing determined by EMG (Crowninshield and Brand 1981). Using EMG to validate models describing joint motion as opposed to hip joint contact forces is common. However, Stansfield et al. (2003) state that, "These instrumented implants must be regarded as the 'gold standard' reference against which all other estimates of hip joint contact force are judged." Unfortunately, based on the four validation studies that have been published, musculoskeletal models have not been fully successfully in validating a model for predicting joint forces (Stansfield et al. 2003; Heller et al. 2001; Brand et al. 1994; Lu et al. 1998). Brand et al. (1994) found resultant hip contact forces from normal and slow gait to be 0.5 BW greater than measured forces using the model developed by Crowninshield and Brand (1981). The calculated out of plane hip contact forces tended to agree fairly well with measured forces although the authors report no metric to define how well the calculated data fit the measured data. The authors note that neglecting the role of knee ligaments and antagonistic muscle activity was the likely cause of the differences between measured and calculated forces. In addition, the authors predicted hip joint contact forces based on motion and external force data measured months after the prosthesis data were collected. Therefore, the model was not based on the exact same movements and forces as the prosthetic data. While using a 2D model, Lu et al. (1998) found good agreement between calculated axial forces in the midshaft of the femur and measured forces when subjects performed isometric contractions. However, the model did not perform as well when predicting femoral midshaft forces during level walking. While the model did predict co-contraction, the two-dimensional nature of the model did

not allow for the contributions of adductors and abductors, limiting the validation study.

Heller et al. (2001) conducted a very detailed study with 4 subjects fitted with

instrumented hip prostheses. Each subject performed several trials of normal walking

and stair climbing. A musculoskeletal model based on anatomy derived from the Visible

Human (NLM, Bethesda, USA) and previous models of Brand et al. (1982) was built and

scaled to represent each of the four subjects. In the first published attempt to validate a

model for cycle-to-cycle prediction of hip contact forces, Heller et al. (2001) found that

the model differed from measured contact forces by 12% during level walking and 14%

during stair climbing. However, after taking care to use detailed anatomy and

sophisticated models, the authors used an optimization routine that had many known

limitations, namely that antagonistic muscle activity (co-contraction), a known

phenomenon, could not be predicted as a possible solution. In a next step, Stansfield et al

(2003) used a similar approach as Heller et al. (2001), but used a different optimization

routine to allow for more muscle recruitment in synergistic muscles as opposed to having

only large muscles contracting. However, the predicted contact forces at the hip were

still 13-18% different than measured forces and, again, no antagonistic muscle activity

was predicted. Thus, while an enormous amount of work has been put into these more

detailed studies, it is hoped that the authors will continue to modify existing models and

optimization routines to eventually validate their measured data. Unfortunately, the

following statement published by Brand et al. in 1982 concerning mathematical methods

in musculoskeletal model creation still holds true: "Every method so far devised is

subject to sufficient limitations so as to question the results and continue the search."

When investigating bone deformation in the human hip, knowing the joint contact forces and muscle forces are crucial. Without valid knowledge of the forces acting at the hip, a valid measure of bone deformation will not be possible. Fortunately, when a model is validated to predict hip joint contact forces many researchers will be ready with models to predict stress patterns using both beam models (Beck et al. 1998; Silva et al. 1999; Mourtada et al. 1996) and finite element models (Ribble et al. 2001; Lotz et al. 1991a, b) of the proximal femur. However, with true validation occurring only when a 3D musculoskeletal model is compared to instrumented hip prostheses, a question arises whether the complexity of these models is really necessary.

Currently, the practical implementation of bone loading protocols in the bone/exercise field does not lend itself to the complexity of musculoskeletal modeling. Furthermore, the implementation of these complex models is extremely difficult to perform, hence the lack of validation studies in the literature. However, while a simple approach to quantifying the intensity of skeletal loading has been presented in this dissertation, any research that serves to increase our understanding into how skeletal loading relates to changes in human bone is a step towards future fracture prevention.

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APPENDIX

INFORMED CONSENT DOCUMENT

Project Title: Local Stresses at the Hip from Drop Landing Exercises in Prepubertal Children
Principal Investigator: Dr. Christine Snow, Oregon State University, Exercise and Sport Science
Research Staff: Jeremy J. Bauer, Dr. Wilson C. Hayes, Dr. Michael Freeman, Dan Edwards

PURPOSE

This is a research study. The purpose of this research study is to determine what stresses drop landings from 24" box place on the hip above normal background activities such as walking, running, and landing from a lower height. The OSU Bone Research Laboratory has demonstrated that children performing drop landings from 24" gain more bone mass at the hip than children not participating in drop landings. The researchers' goal is to answer *why* the differences in bone mass occur. The purpose of this consent form is to give you the information you will need to help you decide whether your child should be in the study or not. Please read the form carefully. You may ask any questions about the research, what you will be asked to do, the possible risks and benefits, your rights as a volunteer, and anything else about the research or this form that is not clear. When all of your questions have been answered, you can decide if you want your child to be in this study or not. This process is called "informed consent". You will be given a copy of this form for your records.

The researchers are inviting your child to participate in this research study because your child is the same age as children who have been in our drop landing intervention studies and changed bone mass as a result.

PROCEDURES

If you agree to allow your child to participate, your child's involvement will last for approximately 3 hours over a period of two days. There will be two testing sessions. The first testing session will take place in the Oregon State University Bone Research Laboratory and will take approximately one hour. The second testing session will take place at the Willamette Spine Center in Salem, OR and will take approximately 2 hours including transportation time. All testing will take place in the period of one week.

The following procedures are involved in this study.

- **Anthropometric Measures:** My child will have her standing height, sitting height and weight measured. My child will also have her foot length, ankle width, knee width, pelvis width, waist height, and shoulder height measured.

- **Health History Questionnaire:** I will assist my child in recording her health history on a questionnaire that will take approximately 5 minutes to complete. The questionnaires will provide current information on health over the previous year.

- **Force and Motion Measures**: First, 30 small, reflective Styrofoam balls will be taped to my child's skin and clothes; loose clothes may be pinned or taped to keep from hiding the balls. We will then tape sensors on your child's left leg to measure when her muscles are active. After a 2-3 minute warm-up of light aerobic activity each participant will perform 15 jumps off a 12", 18" and 24" box onto two force plates (one foot each plate), and jump in place 15

times on the two force plates. In addition, your child will walk and run across the force plates with the goal of measuring 15 foot strikes on the left leg and 15 foot strikes on the right leg for a total of 30 trials running and 30 trials walking. The force plates will record how much force the ground exerts on your child's feet. Six video cameras will be used to record the how your child lands on the force plates. All force and motion data will be collected in one session.

- **E.M.G. (Electromyography):** Measurements of lower extremity muscle activity will be made using surface mounted sensors at the same time force and motion measures are being made. E.M.G. is a noninvasive system used to measure the electrical activity of any muscle. The sensors are placed on the skin much like the reflective markers in "Forces and Motion Measures". However, the skin must first be cleaned with an alcohol swab prior to E.M.G. sensor placement.

- **M.R.I. (Magnetic Resonance Imaging)**: Measurements of lower extremity muscle and bone geometry will be conducted by qualified technicians at the Willamette Spine Center in Salem, OR. Subjects will be transported to the center by either the principal investigator, or the subject's parents. MRI scans will take approximately 45 minutes to complete.

RISKS

The possible risks associated with participating in this research project are as follows.

- **Force and Motion Measures**: No injuries or muscle soreness were incurred during the previous testing in 37 children who performed 100 consecutive drop landings from 24 inches or in 14 children who ran across the force plates in the Bone Research Laboratory. Muscle soreness may occur and will be minimized by the inclusion of a brief warm-up and cool-down. There is also a slight chance of injury due to accident. To minimize accidental injury the area will be free of any obstacles within a 2-meter radius. Trained personnel will closely monitor all exercises performed in the Bone Research Laboratory on the day of testing.

- **E.M.G. (Electromyography):** Surface mounted E.M.G. is noninvasive and poses no risks to the individual wearing the sensors. The sensors are applied and removed similarly to a band-aid or sticker on the skin.

- **M.R.I.**: MRI does not pose any risks to children unless they have any kind of implanted metal objects in the body such as:

  - implanted pacemaker
  - implanted medication device, such as an insulin pump
  - metal clips or pins, or other metal objects in the body
  - any bullet wounds, particularly if the bullet remains in the body
  - any metal joint replacements or heart valve replacements

BENEFITS

There may be no personal benefit for participating in this study. However, the researchers anticipate that, in the future, society may benefit from this study by helping to define an exercise prescription for the prevention of osteoporosis later in life.

COSTS AND COMPENSATION

You will not have any costs for participating in this research project. Your child will be compensated for participating in this research project. If your child completes all of the tests described in the "Procedures" section your child will be paid $100.00. There will be no partial compensation for incomplete tests.

CONFIDENTIALITY

Records of participation in this research project will be kept confidential to the extent permitted by law. However, federal government regulatory agencies and the Oregon State University Institutional Review Board (a committee that reviews and approves research studies) may inspect and copy records pertaining to this research. It is possible that these records could contain information that personally identifies your child. Your child will be assigned a code number, which will be used on all computer output and video tapes and will be stored in a separate file. Only the investigators will have knowledge of your child's name and code number. In the event of any report or publication from this study, your child's identity will not be disclosed. Results will be reported in a summarized manner in such a way that your child cannot be identified.

AUDIO OR VISUAL RECORDING

By initialing in the space provided, you verify that you have been told that visual recordings will be generated during the course of this study. The recordings are being made in order for us to properly estimate how much force the hip is exposed to. Your child will be referenced on the video by the code number described under "Confidentiality". Only the investigators will have access to the videos. The videos will be stored in the Oregon State University Bone Research Lab and will be kept no longer than 3 years.

_____ Parent/Guardian's initials

RESEARCH RELATED INJURY

In the event of research related injury, compensation and medical treatment is not provided by Oregon State University.

VOLUNTARY PARTICIPATION

Taking part in this research study is voluntary. You may choose not to have your child take part at all. If you agree to allow your child to participate in this study, you may stop your child's participating at any time. If you decide not to allow your child to take part, or if your child stops participating at any time, your or your child's decision will not result in any penalty or loss of benefits to which you may otherwise be entitled. All data will be stored as noted in "Confidentiality" unless you would prefer that the data are destroyed.

QUESTIONS

Questions are encouraged. If you have any questions about this research project, please contact: Jeremy Bauer at (541) 737-5935 or by e-mail at bauerje@onid.orst.edu or Christine Snow at (541) 737-6788 or by e-mail at Christine.Snow@orst.edu. If you have questions about your rights as a participant, please contact the Oregon State University Institutional Review Board (IRB) Human Protections Administrator, at (541) 737-3437 or by e-mail at IRB@oregonstate.edu or by mail at 312 Kerr Administration Building, Corvallis, OR 97331-2140.

Your signature indicates that this research study has been explained to you, that your questions have been answered, and that you agree to take part in this study. You will receive a copy of this form.

Participant's Name (printed): _____

_____   _____

(Signature of Parent/Guardian or   (Date)
Legally Authorized Representative)

INVESTIGATOR STATEMENT

I have discussed the above points with the participant or, where appropriate, with the participant's legally authorized representative, using a translator when necessary. It is my opinion that the participant understands the risks, benefits, and procedures involved with participation in this research study.

_____   _____

(Signature of Investigator)   (Date)

ASSENT DOCUMENT

Project Title: Local Stresses at the Hip from Drop Landing Exercises in Prepubertal Children
Principal Investigator: Dr. Christine Snow, Oregon State University, Exercise and Sport Science
Research Staff: Jeremy J. Bauer, Dr. Wilson C. Hayes, Dr. Michael Freeman, Dan Edwards

Researchers at Oregon State University are doing a research study. A research study is a special way to find out about something. The researchers are trying to find out what kind of forces the bones in your hip get when you walk, run, jump, and land. This form is about the study, so you can learn about the study and decide if you want to be in the study or not. You can ask any questions. After all of your questions have been answered, you can decide if you want to be in this study or not.

If you decide that you want to be in this study, we will ask you to do several things:

- **Measure your bones and muscles using a picture**: The researchers will ask you to lie quietly on a table for about 45 minutes so we can take pictures of the bones and muscles in your legs. These measurements will be in Salem, Oregon. This is safe for kids of your age and has been used in many other studies.

- **Measure your height and weight**: The researchers will measure how tall you are when you stand straight, when you sit, how long your foot is, how wide your ankles and knees are, how high your waist and shoulders are, and also how much you weigh.

- **Tell us about your health and how you are growing**: The researchers will ask you (with help if you need it) to write down how healthy you are and also write down what you eat most days. The researchers will also ask you to tell us about whether your body is changing.

- **Jumping, running and walking**:

  We will ask you to:
  - Jump in-place 15 times
  - Jump off a 12 inch tall box 15 times
  - Jump off a 18 inch tall box 15 times
  - Jump off a 24 inch tall box 15 times
  - Walk down a wooden side-walk 30 times
  - Run down a wooden side-walk 30 times

- **Measure your Muscles**: We are going to measure when your muscles are flexing. To measure your muscles we will first clean the skin over your muscles. After cleaning your skin we will tape snap buttons to your skin that will help us measure when your muscles are flexing.

Before you start, the researchers will tape 30 shiny little balls to your skin and your clothes and we will measure the size of your foot, ankle, knee, hips, waist, and shoulder. The researchers are going to measure how hard or soft your feet hit the ground. The researchers will ask you to wear some tight fitting clothing so that we can use video cameras to record how you land. The researchers will ask you to come to the Bone Research Laboratory only one time.

The researchers want to tell you about some things that might happen to you if you are in this study. You might be hurt from the jumping exercises or the walking and running. The risk for injury is small. Other children doing these exercises and tests in school and at Oregon State University have not been injured. Also, you should not get the pictures of your muscles and bones done if a doctor has ever put metal in your body.

These tests will teach you about the muscles and bones in your legs. These tests might also help us find out things that will help other children some day.

When the researchers are done with the study, they will write a report about what they found out. The researchers won't use your name in the report.

You don't have to be in this study. It's up to you. If you say okay now, but you want to stop later, that's okay too. All you have to do is tell the researchers.

If you want to be in this study, please sign your name.

I, _____, want to be in this research study.
      (Print your name here)

_____   _____
      (Sign your name here)   (Date)