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Hip fractures have high mortality and morbidity rates after incidence, with osteoporosis being a major risk factor due to the loss of bone mineral density (BMD). Lower-body resistance exercises, such as squats, can provide sufficient loading on the hip to induce osteogenic effects. However, this loading may depend on how the exercise is performed. The purpose of this study was twofold: determine the loading on the hip as a function of squat depth, and examine the extent to which this relationship is affected by adding static resistance to the upper body. Twenty healthy women, from 35-49 years of age, performed sets of shallow-, medium-shallow-, medium-deep-, and deep-depth squats, both with and without upper-body static resistance in the form of a 5.4 kg weighted vest. From recorded motion capture and ground reaction data, the depth of each squat (i.e. peak knee flexion) and peak loading on the hip joint were calculated using a biomechanical model. Increases in squat depth and in peak trunk flexion increased the overall magnitude of hip loading during the squat exercise by 8.6% body weight (BW)/deg and 2.4% BW/deg, respectively, on average. For squat depths greater than 50°, the weighted vest increased the effect of squat depth on the overall magnitude of hip loading by 2.0% BW/deg. The results suggest that by squatting to deeper depths, with

increased trunk flexion, and while wearing a weighted vest, it is possible to place high loads on the hip that may increase BMD and, in turn, reduce the risk of hip fractures.

Hip Loading During the Squat Exercise

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

Gabriel J. Haberly, Author

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CHAPTER 1 - INTRODUCTION

The occurrence of hip fractures has become more of an issue in recent years. A study found that 1.7 million hip fractures occurred worldwide in 1990 and it is estimated that, by 2050, there will be 6.26 million hip fractures per year worldwide (Cooper et al., 1992). The increasing incidence of hip fractures could create economic problems for individuals, as the cost of hip fractures can be anywhere from $8,358-$32,195 per incident (Budhia et al., 2012). In addition to hospital bills, further funds are spent on skilled nursing care, as some elderly patients do not regain their former level of independence after sustaining a hip fracture (Leibson et al., 2002). The problem of hip fractures is not just the financial costs; the one-year mortality rate after incidence of a hip fracture can be as high as 33% (Roche et al., 2005). In order to provide a means of preventing hip fractures, it is important to identify the risk factors for such fractures. The diagnosis of osteoporosis is an important consideration, as it is highly related to the occurrence of hip fractures (Kanis, 2002). Osteoporosis significantly increases hip fracture risk, as 18% of women who are osteoporotic will experience a hip fracture in their lifetime (Chrischilles et al., 1991).

Originally, a consensus conference defined osteoporosis as “a disease characterized by low bone mass and microarchitectural deterioration of bone tissue, leading to enhanced bone fragility and a consequent increase in fracture risk” (Peck et al., 1993). The cause of osteoporosis can be attributed to many factors, including low peak bone mass, loss of hormones, drug abuse, smoking of cigarettes, insufficient physical activity, calcium and/or vitamin D deficiencies, body size, and genetics (Lane, 2006). Postmenopausal women are more commonly diagnosed with osteoporosis due to the decreased estrogen levels in their body that ultimately lead to a lower bone mineral density (BMD) (Riggs et al., 1998). The cause for concern in individuals with osteoporosis is that, because they have a lower BMD, they are more susceptible to bone fractures (Nevitt et al., 1994).

If clinicians can obtain ways to treat osteoporosis through the prevention of BMD loss, they will, in turn, reduce the risk of hip fractures. The most readily available options for preventing bone loss involve dietary supplements of vitamin D and calcium. These two substances have been shown to help prevent bone loss, as well as maximize the effect of osteoporosis drug therapy (Tang et al., 2007). The use of pharmaceutical drugs can also be applied to osteoporosis to help prevent fractures. It has been shown that nitrogen-containing bisphosphonates have had a positive effect on osteoporosis (Bock and Felsenberg, 2008), but drugs such as these are not available to many people due to large costs. While both supplementations and pharmaceuticals slow the rate of bone loss, they have no effect on fall prevention. This downside to these treatment options decreases their effect on fracture risk, as the primary catalyst for hip fractures is falls (Youm et al., 1999).

Exercise is an alternative treatment, available to most osteoporotic individuals, that can use loading on the musculoskeletal system to address the problems of low BMD. In order to produce the optimal osteogenic effects, the loading of bone during exercise should meet the following criteria: dynamic stimulus; exceed a threshold magnitude; exceed a threshold strain frequency; be relatively brief but intermittent; and impose an unusual loading pattern on the bones (Borer, 2005). Bone can be built through mechanical loading due to either a separate or combined stimulus of external and internal forces. During exercise, external stimuli are associated with ground reaction forces (GRF) and applied resistances forces, whereas internal stimuli are primarily associated with muscle forces. In activities that involve impact with the ground, such as jumping, walking, and running, the GRF is a primary or major mechanism of bone loading (Bassey et al., 1997; Bergmann et al., 1993). In contrast, during “non-impact” activities such as resistance training, biking, and rowing, muscle forces are responsible for the majority of bone loading (Lu et al., 1997). However, both types of exercise are capable of increasing whole body BMD (Kohrt et al., 1997). Beyond its effect on BMD, exercise can also be an effective means of preventing hip fractures by reducing the risk of falls (Shaw and Snow, 1998). All types of exercise may not be appropriate for all individuals, however.

Resistance training can provide maintenance or increases of BMD and increases in muscular strength, aiding in the prevention/treatment of osteoporosis and reducing the risk of falls (Kohrt et al., 1997; Snow and Shaw, 1998), without the possible dangers of high impact forces. The selection of the specific resistance exercises performed is important, however, because different loading stimuli may produce different bone responses (Lanyon and Rubin, 1984). In particular, an individual who is designing an exercise regime must determine which regions of the body an exercise will target and the magnitudes of the loads that exercise places on the targeted regions. To prevent hip fractures, it is important to incorporate exercises that train the lower body, as the response of bone to loading is site specific (Winters-Stone and Snow, 2006). That is, certain lower-body resistance exercises will produce a larger response at the hip than will others.

Additionally, the overall magnitude of loads applied to bone is important, as higher-magnitude loading can result in larger osteogenic effects (Rubin and Lanyon, 1985). If individuals are attempting to increase BMD of the hip, then they should examine lower-body resistance exercises to determine the most appropriate ones to include in their program.

Although exercise programs that include resistance training have been shown to increase BMD in the hip (Going et al., 2003; Kohrt et al., 1997), there has been little research done with regards to the specific internal loading associated with lower body exercises that might create such improvements. In possibly the only study that analyzed hip loading during lower-body resistance training exercises, Anderson et al. (1996) found that the squat exercise produced the highest compression on the femoral neck when compared to hip flexion, hip extension, hip abduction, hip adduction, leg extension, and leg curl exercises. Although that study provides evidence for the importance of the squat exercise in building BMD, it does not give specific information about the technique of the squat exercise that could aid individuals in maximizing hip loading.

When it comes to resistance training, the technique of most exercises is very important in overall safety and development of strength in the individual (Faigenbaum and Myer, 2010). Although some exercises are simple and require little instruction regarding proper technique, the squat does not fall into this category. Different approaches in technique to the squat exercise might result in different loading on the hip joint. One very important aspect of the squat is the maximum knee flexion angle that an individual attains when performing a squat, which is also known as the squat depth. The relationship between squat depth and loading on the hip has not been identified. Although squatting to a deeper depth might produce more loading on the hip, some individuals may have trouble reaching such depths. The addition of static resistance could provide an alternative approach to attaining high hip loading while squatting to shallower depths. An exercise program that included the addition of upper-body static resistance, in the form of a weighted vest, to squat and lunge exercises was found to increase BMD of the hip (Snow et al., 2000). However, the effect of added upper-body static resistance on the relationship between squat depth and hip loading has not been studied. In order to develop the most effective resistance training programs for the prevention of hip fractures, it is important to determine the influences of squat depth and the addition of static resistance on hip loading during the squat exercise.

Problem Statement

The long-term goal of this study is to develop effective exercise programs that will prevent hip fractures. The objective of this study is to examine how the loading on the hip during a squat exercise varies as a function of the depth of the squat, as well as the degree to which the addition of static resistance to the upper body affects loading on the hip at different depths during a squat exercise. The central hypothesis is that loading on the hip will increase as squat depth increases, and that the addition of static resistance will produce greater loading on the hip across a range of depths. The squat exercise has been shown to be a mechanism for loading the hip and potentially increasing BMD, but it can be a difficult task for some individuals. This study will give practitioners further insight to determine the squat depth and static resistance needed to achieve a given loading of the hip during a squat exercise. The central hypothesis will be tested through the following specific aims:

Aim 1: Determine the loading on the hip as a function of squat depth during a squat exercise. It is hypothesized that the loading on the hip will increase as the depth of the squat increases.

Aim 2: Determine the extent to which the addition of static resistance to the upper body affects the relationship between squat depth and hip loading during a squat exercise. It is hypothesized that, across a range of squat depths, performing a squat with added upper-body static resistance will produce greater loading on the hip than performing a squat to the same depth without upper-body static resistance.

Significance

Hip fractures are a worldwide concern and are expected to become more prevalent as the population ages (Cummings and Melton, 2002). Exercise programs have been shown to aid in the prevention of hip fractures by preventing or slowing losses of BMD in the hip (Engelke et al., 2006; Going et al., 2003; Kemmler et al., 2012). More specifically, exercise programs that included lower-body resistance training had a high impact on the prevention of BMD losses in the hip (Winters-Stone and Snow, 2006). Further examination of lower-body resistance exercises would prove helpful in determining how BMD in the hip can be maintained. Specifically, analyzing the squat exercise could prove useful because it produces a greater amount of loading on the hip than many other lower body resistance exercises (Anderson et al., 1996). In order to create effective exercise programs that involve squats, it is important to determine how different approaches to squatting may differ in their effect on BMD. Therefore, this study will determine the relationship of added static resistance and different squat depths to the loading on the hip. The results of this study should give practitioners insight into instructing women on how they should perform the squat exercise in order to achieve beneficial effects on BMD at the hip. When prescribing regimes for individuals to prevent hip fractures, the goal of practitioners should be to ensure safe and constructive exercises that maximize hip loading. Instruction regarding the depth of a squat, and the amount of added static resistance needed to significantly affect loading on the hip, will be helpful in the prevention of hip fractures among individuals of different backgrounds and abilities.

CHAPTER 2 – LITERATURE REVIEW

Purpose and Scope

The purpose of this literature review is to provide background knowledge on the topic of the study that was performed and to provide an understanding of its importance. This review will achieve the following:

- Define the problem of hip fractures, including consequences and risk factors;
- Describe the pathogenesis and diagnosis of osteoporosis;
- Compare and contrast different osteoporosis treatments and interventions and their effects;
- Discuss how exercise creates various mechanical loading which, in turn, stimulates improvements in bone density;
- Examine lower body resistance training as an intervention to improve hip bone mass and prevent fractures;
- Discuss how hip loading can be measured and modeled;
- Conclude with a summary of important take-home messages.

The Problem of Hip Fractures

This study will focus on the squat exercise and how different techniques may aid in the prevention of hip fractures in women. Hip fractures are considered to be the most severe form of osteoporotic fracture, as most patients who suffer a hip fracture are hospitalized (Johnell and Kanis, 2005). The increasing age of the world's population has increased the frequency of hip fractures by 1-3% per year in most areas of the world (Cummings and Melton, 2002). In the United States, lifetime costs of hip fractures have been estimated to be $81,300 per person (Braithwaite et al., 2003), with the total annual cost of hip fractures projected to be $18 billion in 2025 (Burge et al., 2007). Once an individual has sustained a hip fracture, health and quality of life can decrease. Roche et al. (2005) found the following postoperative complications following hip fractures: cardiovascular problems, dementia, respiratory disease, venous thrombosis or pulmonary embolus, deep infection, urinary tract infection, gastrointestinal hemorrhage, and myocardial infarction. In addition, mortality rates after hip fracture can be as high as 33% within one year after the fracture (Roche et al., 2005). Even if no such complications arise, elderly victims of hip fractures will have lifelong physical problems, as the majority do not regain their former level of independence and functionality (Leibson et al., 2002).

The hip consists of a ball-and-socket joint between the femur and pelvis. The proximal end of the femur consists of the head and neck. A key point of interest is the femoral neck, as 90% of hip fractures occur in the shaft (femoral neck fracture) or at the base (intertrochanteric fracture) (Gallagher et al., 1980). Although hip fractures can occur in both trabecular (cancellous) and compact (cortical) bone, it has been shown that fractures occurring in the femoral neck are predominately related to cortical bone loss (Bell et al., 1999). Although cortical bone loss is an important contributor to hip fractures, the root of the problem can come through other means. Decreased estrogen levels can lead to higher endocortical and cancellous bone resorption, resulting in fewer endocortico-trabecular connections (Bagi et al., 1997). The main outcome of this occurrence is weakened bone structure and lesser resistance to outside forces. This can be especially troublesome for women increasing in age, as their natural estrogen levels deplete.

An increased likelihood of falls can be a major risk factor for the occurrence of hip fractures, as falls account for 90% of all hip fractures (Youm et al., 1999). Osteoporosis further increases the likelihood of hip fractures, as 18% of women who are osteoporotic will experience a hip fracture in their lifetime, with a 14% chance of reoccurrence (Chrischilles et al., 1991). Because osteoporosis has been identified as one of the most important risk factors in hip fractures (Kanis, 2002), strategies targeting the prevention of hip fractures should include identifying causes and treatment methods for osteoporosis.

Osteoporosis

Osteoporosis is a disease involving deterioration of the skeleton that leaves bones weak and susceptible to fracture. The cause of osteoporosis can be attributed to both internal and external factors that include low peak bone mass, loss of hormones, drug abuse, smoking of cigarettes, minimal physical activity, calcium and/or vitamin D deficiencies, body size, and genetics (Lane, 2006). Osteoporosis is far more common than and just as dangerous as other diseases that capture public attention (Bock and Felsenberg, 2008). The combined lifetime risks of fractures due to osteoporosis that come to clinical attention are equivalent to the risks of cardiovascular diseases (Kanis, 2002).

At the molecular level, bone is built and remodeled using two different types of cells; osteoblasts (bone building) and osteoclasts (bone absorbing). Osteoporosis takes hold of an individual when the activity of osteoclasts outpaces the activity of osteoblasts

(Lane, 2006). The result of osteoclast takeover creates weaker bones that ultimately lead to skeletal fragility, which in turn increases the risk of fractures. Skeletal fragility can result from a number of factors, such as failure to produce a skeleton of optimal mass and strength during childhood development, excessive bone resorption resulting in decreased BMD along with deterioration of the skeleton, and inadequate formation response to increased resorption during modeling (Raisz, 2005).

In women, the hormone estrogen has been shown to be directly related to the activity of osteoblasts (Tobias and Compston, 1999). As women get older, their production of estrogen is greatly decreased starting around the time of menopause. Rapid decreases in estrogen levels have been shown to create a reduction in both cortical and trabecular bone density (Khosla and Riggs, 2005), which in turn leads to osteoporosis (Riggs et al., 1998). While loss of estrogen is almost impossible to prevent for most women, there are other ways of promoting osteoblast activity and inhibiting osteoclasts, which will be discussed later. This process will greatly reduce the rate of BMD decline.

Osteoporosis is diagnosed clinically by using a dual energy x-ray absorptiometry (DXA) scan that evaluates bone mineral density (BMD) and bone mineral content (BMC). Although this measure provides a way to compare an individual’s BMC and BMD to the general population, it does not always accurately reflect changes in the integrity of bone. An important aspect of continued tests is not just the total BMD and BMC but the recent rate of change of each score. Osteopenia is a precursor to osteoporosis, and is defined as a BMD 1 to 2.5 standard deviations below the mean of 30 year old Caucasian women (Kanis, 2002). Osteoporosis is diagnosed when an individual’s BMD is more than 2.5 standard deviations below the mean of 30 year old woman (Kanis, 2002).

Treatment Methods for Osteoporosis

Pharmaceuticals have increased in popularity recently as a means of treating and preventing osteoporosis. Although not extremely cost effective, osteoporotic drug therapy is a relatively simple process that can provide a means of maintaining bone mass. There are two ways that pharmaceutical treatments directly impact osteoporosis: inhibition of osteoclasts or stimulation of osteoblasts. Estrogens, progesterone, bisphosphonates, and calcitonin have an inhibitory effect on osteoclast activity and bone resorption (Bock and Felsenberg, 2008; Chesnut et al., 2000). Parathyroid Hormone (PTH) and its derivatives have been shown to have a positive effect on bone density (Neer et al., 2001), which is thought to be due primarily to an increase in osteoblast activity (Dobnig and Turner, 1995).

Although pharmaceutical treatment for osteoporosis has shown to be effective in increasing overall BMD, it has its limitations. The limitation that stands out the most is the overall cost of drug therapy. Although many people might be able to receive pharmaceutical therapy for a short-term intervention, sustaining such treatment for a prolonged period of time may prove difficult, as drug therapy may not be cost-effective for all populations (Fleurence et al., 2007). In addition, drug therapies for osteoporosis require full compliance by the patient in order for them to be successful. Compliance for most osteoporotic drug therapies has been shown to decline over time, which in the long-term, would not reduce risks for fractures and would leave patients with high health care costs (McCombs et al., 2004). Even if a patient is compliant throughout the course of therapy, it has been shown that long term exposure (5 years) to estrogen and progestin therapy significantly increases the risk of breast cancer, coronary heart disease, stroke, and pulmonary embolisms (Rossouw et al., 2002). Although pharmaceutical treatment can keep bone mass from reaching osteoporotic levels, it may only be feasible for certain people who can afford the treatment and remain compliant for a sustained period of time.

As bone is mostly composed of minerals, the diet of an individual can have an effect on the overall health of each bone. More specifically, Tang et al. (2007) found that supplementing calcium and vitamin D into the diets of older adults helped reduce the risk of fractures and bone loss. With such findings, one can logically conclude that the supplementation of calcium and vitamin D are an important consideration in osteoporosis prevention programs (Borer, 2005).

Peak bone mass is a factor in osteoporosis prevention that can be influenced with healthy behaviors during the younger stages of an individual's life. Overall peak bone mass is achieved in an individual's early 20's and is maintained throughout the next few decades (Bonjour et al., 1994). If individuals incorporate physical activity into their daily lives at a young age, then they will achieve a higher peak BMD (Gunter et al., 2008). A logical conclusion can be made that if individuals obtain a higher peak bone mass when they are younger, then they will delay the onset of osteoporosis that may occur later in life.

When bone health becomes a concern for individuals, they are most likely past the bone-building phase of their lives and into the prevention-of-bone-loss phase. While some drugs can suppress bone resorption or aid in bone formation, they can be extremely costly and are difficult to be considered a long-term solution. In addition, drugs and supplements only are helpful in aiding bone health, with little influences on other body systems. Because most hip fractures occur after a fall (Youm et al., 1999), it would be logical to create an intervention that targeted both bone health and balance ability of an individual. Exercise is an encouraging treatment option, as it can increase bone health by promoting bone formation, as well as improve musculoskeletal function.

Exercise as a Treatment Method for Osteoporosis

Important Characteristics of Bone Loading and Responses

Bone is a highly vascular and adaptable tissue. At a very simple level, bone follows Wolff's law that is based off of a "use it or lose it" principle (Wolff, 1892). Physical activity will lead to increased bone density in areas where the bone is stressed, and the absence of physical activity will result in a decline of bone mass. Bone is an interesting structure, as it responds differently to differing types of stresses. Research has identified a number of loading characteristics that will influence osteogenic responses in bone.

Based on research using animal models, there appears to exist a relationship between load magnitude, strain rate, frequency of loading, the number of loading cycles on bone, and the time between bouts of loading. In terms of load magnitude, it was shown that when bone was subjected to applied forces over a period of time, forces that resulted in greater peak strain magnitude created greater endosteal bone formation (Rubin and Lanyon, 1985). It has also been shown that, as the strain rate of loading on bone increases, the anabolic response to loading will increase (Mosley and Lanyon, 1998). Bone response to load frequency has also been studied, and it was found that a bone subjected to a low strain rate but at a high frequency (90Hz) exhibited increased trabecular bone volume and thickness compared to a similar strain rate with a low frequency (45Hz) (Judex et al., 2007). Furthermore, the number of cycles to which a bone is loaded is also important, as the bone response increases as the number of cycles increases (Cullen et al., 2001). Even though bone is very good at adapting to different loading stimuli, too frequent bouts of loading can create a decline in the sensitivity to a stimulus. Robling et al. (2001) conducted a study that involved examining recovery times of 0-8 hours between four identical daily loading sessions, with 0.5-14 seconds between loading cycles within a session. They found that only the group that rested for 8 hours between loading sessions restored full mechanosensitivity to bone cells, and that the group that had recovered for 14 seconds between loading cycles had 66-190% greater bone formation than the others. While a greater overall magnitude of bone loading seems to be the best way to elicit an osteogenic response, one must be careful to provide adequate rest time between bouts of loading in order to avoid overload of bone cells. It is important to consider all of these characteristics of bone adaption when creating exercise programs designed for building bone mass.

Effects of Exercise on Bone Response

Exercise has the potential to provide a low-cost, effective way to build and maintain bone mass. In addition, exercise acts in a way of naturally stimulating osteoblasts, as opposed to pharmaceuticals and the side effects associated with drug therapy. Multiple studies and interventions have been done to improve the bone density in specific areas, such as the hip. Engelke et al. (2006) did a 3-year program involving 48 osteopenic women that included low-volume, high-resistance strength training and high-impact aerobics in twice-weekly sessions. Korpelainen et al. (2006) implemented a 30-month impact exercise program involving 84 elderly women with low BMD. Both programs demonstrated a maintenance in BMD of the hip, as compared to a decline in the control group. Korpelainen and colleagues also tracked the effect of their intervention on falls and fractures and found that the exercise group had a lower fall-related fracture count than the control group. This finding suggests that exercise can improve the risk factors of both BMD loss and falls in relation to hip fractures. Exercise has also been shown to have a long-lasting effect on BMD when done at a younger age. Gunter et al. (2008) did a 7-month intervention with 33 children that involved jumping exercises versus a control group of 24 children who performed stretching exercises. After 7 months, the jumpers group had 3.6% more BMC at the hip than did the stretching group. Seven years after the intervention, jumpers still had 1.4% more hip BMC than the stretchers. Exercise can also produce a hormonal response that can aid in bone deposition. Multiple studies have shown that adaptive bone response hormones, such as Growth Hormone (Kraemer, W. J., et al., 1995), Insulin-like growth factor 1 (Rubin et al., 2005), estradiol (Kraemer, R. R., et al., 1995), and PTH (Tsai et al., 1997), surge during acute exercises.

Although exercise programs have been shown to increase and/or sustain BMD of certain areas of the body, it is important to determine what types of exercises create such improvements. The loads created on bone by exercise will differ between exercises and parts of the body. A multitude of external ground reaction forces (GRF) and internal muscle forces can be produced during exercise, and the total load acting on the hip will be determined by the combination of both types of forces. The extent to which each type of force contributes to the loading of the hip, and the magnitude and direction of this loading, will depend largely on the task. Different tasks will not just produce different types of loading; they will also produce different magnitudes of loading on the hip (Bergmann et al., 2001). The combination of external forces and internal muscle activity can provide an effective means to load the hip joint that may produce an osteogenic effect.

Impact exercise training that creates high-magnitude loads on the lower body has been shown to be an effective way to increase BMD of the hip (Korpelainen et al., 2006). However, exercise by resistance training can also provide effective increases in BMD and can be more appropriate for older populations (Kerr et al., 1996). In addition to being compatible with multiple populations, resistance training has a larger effect on lean body mass and strength than do impact exercises (Kohrt et al., 1997). This aspect of resistance training can provide a means to increase balance and coordination and decrease the risk of falls (Shaw and Snow, 1998).

Although exercise is an effective means of preventing hip fractures, all types of osteogenic exercises may not be appropriate for all groups of individuals. Sports and plyometrics can create large GRF that have great potential to increase peak bone mass. However this type of training and exercise is much more appropriate for younger adults. In regards to an older female population, high-impact exercise may produce an overload of muscles and bones resulting in more harm than anything else. Resistance training can potentially produce similar high loading on bone through both muscle forces and GRF that will accomplish similar results and help prevent hip fractures.

Resistance Training for Hip Osteoporosis

Resistance training is a type of exercise that is appropriate for most populations looking to increase muscular strength and coordination. Women who are susceptible to osteoporosis may use resistance training to get increases in BMD that may help prevent hip fractures. In older women, lower-body exercises have been found to increase BMD in site-specific areas, such as the hip (Winters-Stone and Snow, 2006). Thus, women interested in increasing BMD of the hip should use resistance training exercises that specifically target the hip. Lower-body exercises will be the most effective resistance training in increasing BMD of the hip, as increases in lower-body strength have been shown to be positively correlated with increases in BMD of the hip (Kerr et al., 1996). One exercise that has been a part of exercise programs that have developed lower-body strength and shown increases in BMD of the hip is the squat exercise (Kohrt et al., 1997). However, there is an apparent gap in knowledge when it comes to different approaches to performing the squat exercise and how they contribute to the loading on the hip.

The Squat Exercise and Hip Loading

Many lower body exercises are available to individuals who seek to create loading on the hip. However, interventions should be as effective as possible and seek to incorporate the best available exercise options. The squat exercise can be found in almost any intervention program that involves lower-body resistance training, as the associated muscle activation patterns are effective in creating large dynamic loads in the hip and lower body. Anderson et al. (1996) studied loading on the hip with seven basic lower-body exercises that included hip flexion, hip extension, hip abduction, hip adduction, leg extension, leg curls, and squats. When all seven exercises were compared, the squat exercise produced the most loading on the hip. A conclusion can be made that squats are one of the most important resistance exercises for increasing BMD of the hip, and thus should be used in hip fracture interventions.

Although the squat itself is not a complex task, it can prove to be difficult for some individuals to perform correctly. The technique of the squat is very important in order to maintain safe musculoskeletal loading and obtain the proper activation of particular muscle groups (Faigenbaum and Myer, 2010). Differing trunk angles (McLaughlin et al., 1978), knee anterior positions (Fry et al., 2003), and maximum knee angles (Bryanton et al., 2012) can create a multitude of different muscle forces acting on the body during a squat. For appropriate muscle activation in the squat exercise, it has been suggested that the feet should be positioned just past shoulder width, the lumbar spine should remain erect, the knees should move slightly past the toes during descent, and the individual's gaze should be directed forward or upward (Comfort and Kasim, 2007; Fry et al., 2003; McKean et al., 2010).

Squat depth is another potentially important factor with regards to musculoskeletal loading during a squat. As squat depth increases, the total number of activated muscles remains relatively constant, but the gluteus maximus becomes more active in concentric contraction (Caterisano et al., 2002). It is possible that this increased muscle activation will lead to an increase in hip loading during a squat, but such a study has yet to be conducted.

There is a need to quantify the effects of squat technique on hip loading. As noted earlier, increased loading on bone can stimulate bone deposition that will eventually lead to an increase in BMD. However, it is important that exercises being used for this purpose be executed in such a way that the amount of loading created will produce a meaningful osteogenic reaction. Yet, when developing intervention programs for older adults, avoiding the generation of too much force may be a key factor to limiting possible side effects. An ability to adapt the exercises being used to age-related reductions in functional capabilities is also desirable. A simple act of using a weight vest during lower-body resistance exercises can provide a means of reducing falls risk (Shaw and Snow, 1998) and possibly maintaining hip BMD (Snow et al., 2000). However, little is known of how the added resistance will affect the loading on the hip during specific lower-body resistance exercises, such as squats. Thus, it is logical to assess the effects of upper-body static resistance (weighted vest) during the squat exercise to determine the possible benefits and gains.

Modeling Hip Loading

Few techniques have been developed to directly measure the mechanical loading of bone in vivo. Loading on bone is difficult to measure because bone is located deep within the body and the loading on bone is constantly changing. Some studies have used strain gages to infer bone loading by measuring bone strain. For example, Aamodt et al. (1997) used strain gages to measure proximal and lateral femoral strains during single-leg stance, double-leg stance, walking, and stair climbing. Studies have also combined multiple measuring techniques, such as using strain gages with a finite element analysis and digital image correlation, to estimate bone strain and compression (Sztefek et al., 2010). Other studies have used instrumented implants to measure forces acting on the hip during daily activities such as walking, stair climbing, knee bends, and rising from a chair (Bergmann et al., 2001), as well as during jumping exercises and running (Bassey et al., 1997). While these techniques are extremely accurate, they are not cost- or time-effective for the current study. A much easier way to calculate the loading on bone during a squat would be the use of a biomechanical model. If a biomechanical model was developed to estimate loads placed on the hip during squats, the effects of different techniques to the squat exercise on mechanical loading of the hip could be determined. Many models have been created that can use measured ground reaction forces, kinematics, and/or electromyography (EMG) recordings to predict the forces acting at a joint (Callaghan and McGill, 2001; Freivalds et al., 1984; Glitsch and Baumann, 1997). In predicting forces acting on the hip, both the external forces acting on the modeled body segments and the muscle forces acting across the joint need to be measured or estimated. In order to estimate muscle forces, techniques involving measured EMG activity (Callaghan and McGill, 2001) or optimization (Glitsch and Baumann, 1997) through the use of a biomechanical model are often used.

Like most computer-based simulations and systems, biomechanical models do have their sources of error. The optimization procedure performed by a biomechanical model is one source of error. This procedure uses inverse dynamics to calculate angular velocities and acceleration, inertial moments and forces, and reactive moments and forces at each joint (Wehner et al., 2010). During the inverse dynamic analysis, the muscle recruitment pattern is optimized to the motion of the task in order to solve for the muscle and joint contact forces. Errors in estimating muscle forces can occur due to the muscle geometry and differing activation patterns. The underestimation of muscle coactivation is another source of optimization error that biomechanical models cannot avoid. In order to account for coactivation error, EMG-assisted models are necessary. Fortunately, Sousa et al. (2007) did a study that analyzed EMG activities of the rectus femoris, biceps femoris, tibialis anterior, and soleus muscles during squats performed to 40, 60, and 90 degrees of knee flexion with both a flexed and straight trunk. It was found that only when the subjects squatted at 40 degrees of knee flexion with a flexed trunk was coactivation of the rectus femoris and biceps femoris present (Sousa et al., 2007). Therefore an EMG-assisted model will not be used in this study, as it would complicate the procedures and is not completely necessary. Instead, the AnyBody Modeling System (AnyBody Technology, Aalborg, Denmark) will be used.

The AnyBody Modeling System uses motion capture and ground reaction force data as boundary conditions for an inverse dynamics and optimization approach. The software first optimizes the dimensions, marker locations, and motion of the model to best fit the motion capture data. Once this is completed, the model performs an inverse dynamics and static optimization procedure to compute the muscles forces and joint contact forces acting within the model during a given trial. The biomechanical model chosen for this study is a modified version of a detailed rigid body model of the hip, thigh, and lower leg from the AnyBody Managed Model Repository. The body model to be used is a three-dimensional model that consists of 11 rigid segments (head-trunk-and-pelvis, thighs, shanks, patellas, tali, feet) with 24 degrees of freedom and 55 muscles divided into 169 fascicles on each leg. Each muscle will be a three-element Hill-type muscle model consisting of a contractile element, a parallel-elastic element, and a series-elastic element. The contractile element includes the force-length relationship, force-velocity relationship, and the pennation angle of the muscle fibers. The parallel-elastic element models the passive force-length relationship of the muscle using a nonlinear spring. The series-elastic element reproduces the nonlinear-elastic behavior of the tendon. The properties of each muscle are determined by the following parameters: peak isometric force, optimal fiber length, tendon slack length, tendon strain at peak isometric force, pennation angle at optimal fiber length, fraction of fast twitch fibers, peak contractile velocity of each fiber type, series-elastic and parallel-elastic force-deformation shape factors, and flexibility of the parallel-elastic element.

Damsgaard et al. (2006) reviewed the AnyBody Modeling System with regards to the effectiveness of its inverse dynamics analysis. It was concluded that the AnyBody Modeling System delivers an accurate analysis when analyzing relatively slow-motion tasks. Several other studies have used the AnyBody Modeling System to determine muscle activation and bone loading during dynamic tasks and have validated the results. Wehner et al. (2010) used AnyBody to calculate internal forces and moments acting on the femur during the gait of a rat. The resulting muscle activation patterns of the vastus lateralis, biceps femoris, and gastrocnemius were compared to the literature and found to be in agreement. In a different study, Wehner et al. (2009) used AnyBody to calculate the hip contact force, as well as the axial force on the tibial plateau, during human gait. The calculated hip contact force and axial force on the tibial plateau were found to be in agreement with the literature of in vivo forces measured by telemeterized joint replacements during gait. Relative to the current study, the squat exercise is a relatively slow-motion task and has a similar relative speed of motion as the human gait tracked in the study conducted by Wehner et al. (2009). Thus, it would appear that the AnyBody Modeling System will provide a valid approach to estimating hip loading during the squat exercise.

Summary

Hip fractures can prove to be a great burden, with increased financial costs, morbidity, and even mortality in women. Prevention and treatment of osteoporosis by minimizing bone loss can be an important factor in reducing the risk of hip fractures. Using pharmaceuticals to treat or prevent osteoporosis is costly and may produce negative long term effects. Exercise in the form of resistance training is an alternative treatment that can reduce the risk of hip fracture through increased BMD and muscular strength. Bone deposition is directly related to the frequency and magnitude of external and internal stimulus. Repeated loading of bone over time can lead to an increase in BMD in certain areas of the lower body. Specific resistance training interventions can be developed to target BMD in the hip joint and thus reduce the risk of hip fracture. In order to optimize such exercise programs, research must be done to identify specifications of certain exercises that will produce sufficient loading on the hip. A better understanding of the manner in which different approaches to the squat exercise affect the loading of the hip will allow practitioners to develop the most effective training programs to prevent hip fractures.

CHAPTER 3 - METHODS

Participants

The participants of this study consisted of 20 healthy women aged 35-49 (mean ± SD age: 42.9 ± 4.8 years, height: 167.8 ± 5.0 cm, mass: 65.5 ± 9.1 kg). In order to be included in the study, participants had to be from 35-50 years of age and must have participated in 20 minutes or more of moderate-to-high intensity physical activity on at least two days per week for each of the previous four weeks. Moderate-to-high intensity activities include those such as strength training, yoga/Pilates, aerobics, dance, swimming, bicycling, and running. The exclusion criteria for the study were the following:

1. If participants had a past or present injury or condition that would presently make squatting difficult or painful;
2. If participants ever had any of the following: chronic low back pain, serious back injury, or surgery on the back, hip, or knee;
3. If participants had any of the following in the previous six months: balance problems or dizziness, back pain, a broken bone in the lower limb, a head injury, concussion, or loss of consciousness, surgery, or pregnancy;
4. If participants presently had any of the following: osteoporosis or bone disease, neurological problems or conditions, a heart or lung problem that limited the ability to exercise, or cold, flu, or sinus symptoms;
5. If participants had taken any of the following types of drugs or medications in the past 24 hours: alcohol (2 or more drinks), sedatives or anxiety relief

medication, recreational drugs, antihistamines, anti-inflammatory medication, or pain relievers.

A screening questionnaire was developed and used to determine whether a participant could be appropriately included in the study (Appendix B). Institutional Review Board approval of this study was obtained and all participants provided written informed consent (Appendix A) before they engaged in any of the study activities.

Instruments and Apparatus

For estimation of the forces on the hip joint during a squat exercise, kinematic and ground reaction force data were collected and used for calculations in a modified version of the Twente Lower Extremity Model (AnyBody Technology, Aalborg, Denmark). Kinematic data were measured at 60 Hz using a nine-camera motion capture system (Vicon, Los Angeles, CA). Simultaneously, ground reaction forces beneath each foot were sampled at 360 Hz from two force plates (Bertec, Columbus, OH).

Procedures

Participants came to the OSU Biomechanics Laboratory for testing. When participants arrived, their informed consent was obtained. They then completed the inclusion/exclusion criteria questionnaire and the research staff evaluated the answers given to determine whether the participant was to be included in the study. Individuals who did not meet the requirements to be in the study were withdrawn from the study.

After participants were admitted to the study, they changed into clothes appropriate for exercise. Participants either wore spandex shorts or shorts of mid-thigh length or shorter during the study. When participants had changed into appropriate clothing, their thigh length was measured using a measuring tape. Participants then performed a task-specific warm-up that allowed for muscle preparation and practice of the squat exercise at different depths. The research staff gave instructions and demonstrations during the warm-up to ensure that participants were completing each task appropriately. The warm-up consisted of the following:

1. A three-minute walk;
2. A set of three body-weight squats at deep depth, or until the first correctly-performed squat after six attempts;
3. One body-weight squat at shallow depth;
4. One body-weight squat at medium-deep depth;
5. One body-weight squat at medium-shallow depth;
6. Repeat of steps 2-5 while wearing a weighted vest (5.4 kg).

The four squat depths were determined based on the peak knee flexion angle. A shallow-depth squat corresponded to a position in which the knees were above the toes and the hips were above the heels. For the deep-depth squat, participants were instructed to perform a squat with as much knee flexion as they felt they safely could, without their hips going lower than their knees and without touching a chair that was located behind them. The research staff estimated the peak knee flexion during the shallow- and deep-depth squats of each participant during the warm-up. The designated squat depths for the medium-shallow and medium-deep squats corresponded to a peak knee flexion angle that was approximately one-third and two-thirds of the way, respectively, between the peak knee flexion seen during the shallow-depth squat and during the deep-depth squats.

Regardless of the depth of the squat, participants were instructed to follow a set of guidelines designed to ensure proper technique. In the initial standing position, the participant was to have her feet slightly wider than shoulder-width apart with weight equally distributed on each foot, and the arms of the participant were to be folded across her chest with hands on opposite shoulders. In performing the squat, the eccentric phase was to begin with the participant flexing the knees and the hips simultaneously, thereby lowering the pelvis until she reached the desired squat depth. Then, without pausing, the participant was to begin the concentric phase and extend the knees and hips to return to an upright standing position. During both phases, the participant was to keep her weight on the posterior portion of the feet. In addition, the participant was instructed to keep her knees and shoulders from moving anterior to the tips of the toes. The spine was to remain in a neutral alignment during the entire motion of the squat. Each participant was instructed to keep her head up, with eyes fixed on a location directly in front of her.

During the warm-up, the timing of the deep-depth squats was controlled using a series of beeps generated 1 second apart by a simulated metronome. Upon hearing the first beep, each participant began the downward motion of the squat; she reached the lowest point of the squat at the second beep, and returned to the upright standing position (i.e. finished the squat exercise) at the third beep. Participants were instructed to perform the remaining squats at this same average speed, both during the warm-up and during the experimental trials.

For the warm-up only, a mirror was set up in front of the participants in order to provide visual feedback regarding their technique. If, during the warm-up, participants did not perform a squat correctly, they were given feedback and asked to repeat it.

Participants were limited to a total of 10 squats in each set at deep depth and three squats in each set at each of the other depths during the warm-up. If, for either set of deep-depth squats, a participant had been unable to complete at least one squat correctly, she would have been withdrawn from the study.

In order to ensure proper safety during the warm-up and experimental trials, a chair with a seat height of approximately 45 cm was set up directly behind each participant, at a distance of approximately half of her thigh length, as she performed the squats at each depth. The chair would have provided support if the participant lost her balance and began to fall backwards.

Once a participant successfully completed the warm-up, she had a set of 24 reflective markers attached to her skin or clothing. The marker locations included:

- Trunk: six markers (clavicular notch; C7 vertebra; right scapula; and left, middle, and right torso on an elastic strap attached at the T10 vertebra)
- Pelvis: six markers (right and left anterior superior iliac spine; right and left posterior superior iliac spine; and right and left lateral pelvis)
- Legs: four markers each (lateral thigh, lateral femoral epicondyle, lateral shank, and lateral malleolus, with the thigh and shank markers offset from their base by a wand of approximately 5.5 cm)
- Feet: two markers each (heel and 2nd metatarsal)

The participant then proceeded to the testing. Each participant performed one set of squats with and one set of squats without added resistance in the form of a 5.4 kg weighted vest (CAP Barbell, Houston, TX). Prior to each set of squats, the participant performed a static trial in which she was filmed while standing in a known reference position. If there were any problems with the static trial, another trial was performed until a correct trial was obtained. Once the static trial(s) was completed, participants then performed squats at deep depth, medium-deep depth, medium-shallow depth, and shallow depth. Each set of squats nominally contained three trials at each squat depth. However, a fourth trial was done at a given squat depth if, during any of the first three trials at that depth, a participant did not perform the correct squat technique (any notable deviation from the technique previously described) or if it was judged that her peak knee angle did not approximate the designated angle. Trials were blocked in such a way that participants performed all repetitions at the current depth before moving on to the next depth. The order in which the blocks were performed within a set was counterbalanced across participants. It was randomly determined whether a participant performed the set of squats with or without the weighted vest first. Participants did not change the static resistance condition (i.e. no weighted vest vs. weighted vest) until they completed the set at their current static resistance condition.

For each experimental trial, participants stood with each foot on a different force plate. The research staff first instructed the participant to place her feet in the proper location on the force plates for the squatting exercise. Then, masking tape was placed on the force plate to mark the participant’s foot positions so that she could maintain the same position throughout the study.

Before the first trial at each squat depth, a research staff member modeled the current squat depth that participants were to perform. The participant then performed the corresponding squat trials at that depth. Motion capture and ground reaction force data were collected during each squat trial. Also, the squat technique that each participant used was continually monitored and the research staff gave feedback to each participant after each trial regarding the squat depth and technique. After participants had completed the required number of trials at any given depth, they then proceeded to the next depth in the sequence. A brief rest period was provided between trials and sets.

After the experimental trials were concluded, the research staff measured each participant's body height using a stadiometer (Seca, Hamburg, Germany) and body weight using a standard scale (Tanita, Tokyo, Japan). An anthropometer (Lafayette Instrument, Lafayette, IN) was also used to obtain the following anthropometric measurements: ankle width, knee width, and foot length.

Data Analysis

Biomechanical Model

The AnyBody Modeling System (AnyBody Technology, Aalborg, Denmark) calculates joint angles and joint contact forces, muscle forces, and joint moments acting on the body. AnyBody has created a standard lower body model called the Twente Lower Extremity Model, which was built using the AnyBody software and is available through their Managed Model Repository version 1.2. A modified version of this lower body model was used. The main modification was to change the knee from a hinge joint to a ball-and-socket joint. Additionally, passive joint torques were added to the knee joint in order to provide passive restraint to adduction/abduction and internal/external rotational motion. These torques were each modeled by a linear torsional spring, with stiffness as determined from Markolf et al. (1981). A cylindrical wrapping surface was added to the proximal femur in order to prevent the gluteus maximus from passing through the femur at large hip flexion angles. Finally, segment masses and inertial properties were modified based on de Leva (1996).

The modified lower body model is a three-dimensional model that consists of 11 rigid segments (head-trunk-and-pelvis, thighs, shanks, patellas, tali, feet) with 24 degrees of freedom and 55 muscles divided into 169 fascicles on each leg. Each muscle is a three-element Hill-type muscle model consisting of a contractile element, a parallel-elastic element, and a series-elastic element. The contractile element includes the force-length relationship, force-velocity relationship, and the pennation angle of the muscle fibers. The parallel-elastic element models the passive force-length relationship of the muscle using a nonlinear spring. The series-elastic element reproduces the nonlinear-elastic behavior of the tendon. The properties of each muscle were determined by the following parameters: peak isometric force, optimal fiber length, tendon slack length, tendon strain at peak isometric force, pennation angle at optimal fiber length, fraction of fast twitch fibers, peak contractile velocity of each fiber type, series-elastic and parallel-elastic force-deformation shape factors, and flexibility of the parallel-elastic element.

The AnyBody software scaled the model to each individual. Segment lengths and pelvis width were either scaled to body height, calculated from the motion capture data, or obtained from direct measurements. The mass of each segment was scaled to the body mass of the individual; the center of mass location of each segment was scaled to the segment length, and the mass moments of inertia of each segment were computed from the mass and length of the segment. The musculoskeletal geometry of each individual was also scaled using a geometric scaling function in which dimensions along and perpendicular to the longitudinal axis were scaled to segment length and $\sqrt{\text{mass/length}}$, respectively. Finally, the AnyBody software scaled the strength of each individual based on the segment mass per length and an estimated percentage of body fat.

The AnyBody software used motion capture and ground reaction force data as boundary conditions for an inverse dynamics and optimization approach. The software first optimized the dimensions, marker locations, and motion of the model to best fit the motion capture data. The tendon slack lengths of the muscles in the model were then calibrated by placing joints in positions corresponding to each muscle's individual optimal fiber length and then adjusting the tendon slack lengths accordingly. Finally, the model performed an inverse dynamics procedure to compute the muscles forces and joint contact forces acting within the model during a given trial. To determine individual muscle forces, the following cost function was minimized for each frame of data in the optimization procedure:

$$J = \sum \left(\frac{}{}\right) \sum \sum$$

where = force generated in muscle *i*, and = maximum force capacity of muscle *i*.

Calculation of Peak Trunk Angle, Knee Angle, and Hip Loading

Before motion capture data and ground reaction force data were input into the AnyBody model, they were filtered using a fourth-order, zero-lag Butterworth low-pass filter with a cut-off frequency of 6Hz and 30Hz, respectively. A BodyBuilder (Vicon, Los Angeles, CA) kinematic model was used to compute joint center locations, body segment dimensions, and initial estimates of the positions of each marker relative to its body segment during the first completed static trial for input to the AnyBody modeling software. The BodyBuilder kinematic model was also used to compute the participant's three-dimensional trunk angles over the course of each trial, based on the positions of the markers attached to the trunk. Calculations of trunk angle were based on the following sequence of Cardan rotations of the trunk relative to the global reference frame: flexion-extension about the mediolateral axis, lateral bending about the anteroposterior axis, and axial rotation about the proximodistal axis. The average orientation of the trunk during the static trial without the weighted vest was used as the reference (i.e. zero) orientation.

The filtered motion capture and ground reaction force data for each trial, as well as the parameters extracted from the static trial, were inserted into the AnyBody model to compute the joint contact force acting on the femur at each hip over the course of the trial. The continuous loading (i.e. joint contact force) on the femur at each hip joint in each direction (i.e. proximodistal, mediolateral, anteroposterior) relative to the femur was then determined by the AnyBody model, as was the flexion angle of each knee. Coordinate directions for the femur were defined based upon the International Society of Biomechanics recommendations (Wu et al., 2002). The proximodistal axis was the line through the hip joint center and the midpoint between the two femoral epicondyles; the mediolateral axis was perpendicular to the proximodistal axis and lay in the plane defined by the hip joint center and the two femoral epicondyles, and the anteroposterior axis was perpendicular to the proximodistal and mediolateral axes. Knee flexion angle was calculated based on the following sequence of Cardan rotations of the shank relative to the thigh segment: flexion-extension about the mediolateral axis, abduction-adduction about anteroposterior axis, and internal-external rotation about the proximodistal axis. A custom MATLAB program (MathWorks, Natick, MA) computed, from the model output, the peak forces acting on the femur in the lateral, distal, and posterior directions at each hip joint, the peak overall magnitude of the joint contact force at each hip joint, and the peak flexion angle of each knee. Peak hip loading was only calculated during the squatting action of the trial. That is, if higher loads were observed at the beginning and/or end of the trial (i.e. in the standing position), they were not included in the analysis. Another MATLAB program calculated the peak trunk flexion angle during each trial from the output of the BodyBuilder kinematic model. Peak hip forces and knee flexion angles were averaged between the right and left limbs for each trial. Averaged peak forces of loading were normalized to body weight (BW) for all participants.

Statistics

Mixed-model linear regression was used to determine the relationship between squat depth, peak trunk flexion angle, the use of the weighted vest, and four different dependent variables: peak loading of the femur at the hip in three directions (distal, lateral, posterior) relative to the femur and the overall magnitude of hip loading. Squat depth was quantified as peak knee flexion angle minus 70º, which corresponded to the average value across trials. A dichotomous indicator variable was incorporated into the regression model to determine the effects on peak hip loading associated with adding static resistance (i.e. the weighted vest) to the squat exercise. Two random- effects variables were used to categorize possible variations between participants. The mixed regression model had the form:

$$= \quad + \quad + \quad + \quad \Delta \quad + \sum \quad + \sum \quad + \quad \Delta \sum \quad + \quad \Delta \sum \quad + \sum$$
$$+ \sum .$$

This equation had the following parameters:

= Peak hip loading of the ith participant during the jth trial

= Participant-specific random effect

$\Delta \sum$ = Squat depth of the ith participant during the jth trial

= Static resistance indicator variable (0 = No resistance, 1 = Resistance)

= Peak trunk flexion angle of the ith participant during the jth trial

= Intercept for 70º knee flexion and 0° trunk flexion, with no static resistance

= Slope coefficients

= Error

In the regression model, and the were fixed-effect coefficients, whereas and were random effects. The correlation between peak knee angle and peak trunk angle was also computed using linear regression.

The data for all valid trials (i.e. judged during data collection to have been performed with the proper technique) were included in the analysis. A range of one to three trials at each squat depth was included for each participant for each static resistance condition. However, one participant was dropped from the analysis because the AnyBody model calculations yielded highly asymmetrical loading between the right and left hip for the majority of her trials without noticeable asymmetries in kinematics or ground reaction forces between legs. A total of 441 trials from 19 participants was thus included in the statistical analyses.

Using a multiple regression power analysis (Cohen et al., 2003), it was determined that a sample size of 20 participants would be able to detect a correlation of $r = 0.55$ in the relationship between squat depth and peak hip loading with a power of 0.8. The level of statistical significance was set to $\alpha = 0.05$ in all analyses and statistical analyses were performed using SPSS version 21 (IBM, Armonk, NY).

CHAPTER 4 – RESULTS

The AnyBody-computed directional (i.e. distal, lateral, posterior) loading of the femur at the hip and the overall magnitude of loading during the squats at each depth followed a similar pattern across participants. For the medium-shallow, medium-deep, and deep squats, all three directional loads relative to the femur tended to show an initial decrease and they then increased as participants began the squatting action; the loading reached a peak either at the time of peak knee flexion or shortly thereafter, and then decreased as participants returned to standing (Figure 4.1). For the shallow squats, it was not uncommon to see higher computed loads at the beginning and/or end of the trial than at the halfway point. However, the loading followed a similar pattern to the other depths, with a peak occurring at the time of peak knee flexion or shortly thereafter (Figure 4.1). It was this peak that was included in the analyses. Loading patterns for participants were similar between trials with and without the 5.4 kg weighted vest across all depths. Across participants and conditions, peak trunk flexion during the squat trials was correlated to peak knee flexion; as peak knee flexion increased, so did peak trunk flexion (r = 0.66).

Based on the derived mixed-effects regression model, the peak force acting on the femur in the distal direction at the hip was influenced by the peak knee flexion angle, the peak trunk flexion angle, and use of the weighted vest (Table 4.1, Figure 4.2). The predicted peak distal force in the reference condition (i.e. 70° of knee flexion, 0° of trunk flexion, no weighted vest) was 421.0% BW. Peak distal force on the femur increased with greater peak knee flexion, greater peak trunk flexion, and, at 70° of knee flexion, with use of the weighted vest. In addition, significant interaction effects indicated that the effect of the vest on peak distal force increased with increasing knee flexion, whereas the effect of increased knee flexion decreased with increasing trunk flexion. The effect of the vest did not depend explicitly on peak trunk flexion (p = 0.497). The mixed-effects regression model explained 94% of the overall variance in the peak force in the distal direction.

The peak force acting on the femur in the lateral direction at the hip was influenced by the peak knee flexion angle, the peak trunk flexion angle, and use of the weighted vest (Table 4.1, Figure 4.3). The predicted peak lateral force in the reference condition was 150.6% BW. Peak lateral force on the femur increased with greater peak knee flexion and greater peak trunk flexion. In addition, a significant interaction effect indicated that the effect of increased knee flexion decreased with increasing trunk flexion. There was no significant effect of the weighted vest on peak force in the lateral direction at 70° of knee flexion (p = 0.157); however, across knee angles, the vest had the effect of increasing the peak lateral force with increasing knee flexion. The effect of the vest did not depend explicitly on peak trunk flexion (p = 0.281). The mixed-effects regression model explained 94% of the overall variance in the peak force in the lateral direction.

The peak force acting on the femur in the posterior direction at the hip was influenced by the peak knee flexion angle and peak trunk flexion angle (Table 4.1, Figure 4.4). The predicted peak posterior force in the reference condition was 33.7% BW. Peak posterior force on the femur increased with greater peak knee flexion and greater peak trunk flexion. A significant interaction effect also indicated that the effect of increased knee flexion increased with increasing trunk flexion. There was no significant effect of the weighted vest on peak force in the posterior direction at 70° of knee flexion (p =

0.770), and the effect of the vest did not depend explicitly on either peak knee flexion (p = 0.296) or peak trunk flexion (p = 0.125). The mixed-effects regression model explained 88% of the overall variance in the peak force in the posterior direction.

The overall magnitude of the peak force acting on the femur at the hip was influenced by the peak knee flexion angle, the peak trunk flexion angle, and use of the weighted vest in the same manner as was the peak distal force (Table 4.1, Figure 4.5). The predicted peak force magnitude in the reference condition was 440.6% BW. As for the peak distal force, the peak force magnitude increased with greater peak knee and/or trunk flexion, and the effect of increased knee flexion decreased with increasing trunk flexion. The weighted vest acted to increase the magnitude of the peak force at 70° of knee flexion, and this effect increased with increasing knee flexion but did not depend explicitly on peak trunk flexion (p = 0.746). As a result, there is little difference in the peak force magnitude at the hip between squats with and without the weighted vest for peak knee flexion angles ranging from 35 to 55 degrees (Figure 4.5). However, as peak knee flexion increases from 50 to 105 degrees, the vest gradually increases the peak magnitude of the force acting at the hip relative to squats performed without a vest. Specifically, the predicted difference in peak force magnitude between the use versus non-use of the weighted vest is 2.1, 42.1, and 82.1% BW at knee flexion angles of 50, 70, and 90°, respectively. The mixed-effects regression model explained 94% of the overall variance in the magnitude of the peak force.

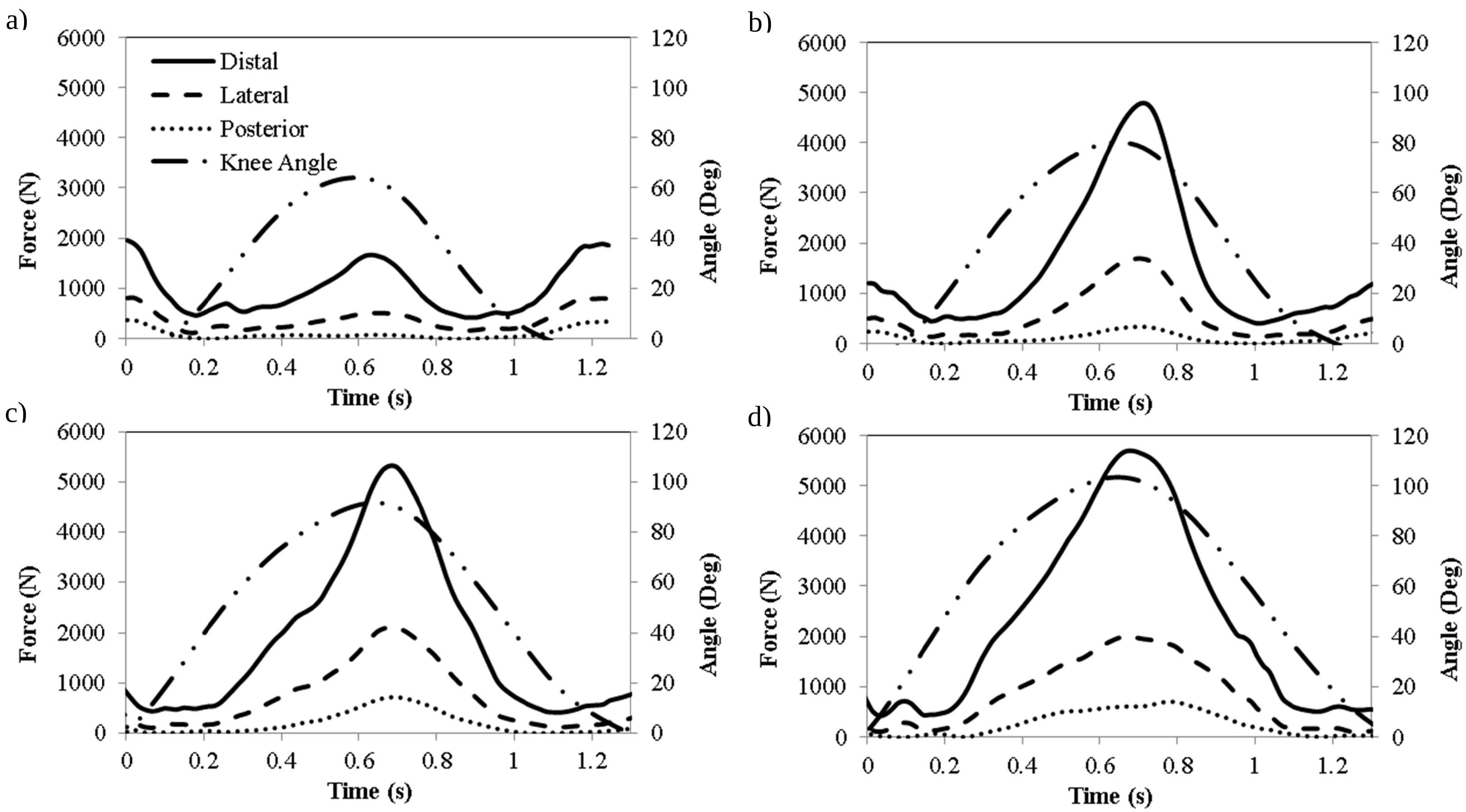

Figure 4.1: Computed directional loading on the femur at the hip and the corresponding knee flexion angle at each time point for a single, 65.6 kg participant during representative trials performed without a weighted vest for the a) shallow squat, b) medium-shallow squat, c) medium-deep squat, and d) deep squat. All loading directions (distal, lateral, posterior) are relative to the femur.

Table 4.1: Mean ± standard error regression coefficients to predict the peak force on the femur at the hip in the distal, lateral, and posterior directions and the peak overall force magnitude.

Variable	(units)	Distal	Lateral	Posterior	Magnitude
Intercept	(%BW)	421.0 ± 33.2‡	150.6 ± 13.7‡	33.7 ± 4.9‡	440.6 ± 35.2‡
$\Delta\theta_k$	(%BW/deg)	12.2 ± 0.9‡	4.2 ± 0.4‡	0.69 ± 0.16‡	12.7 ± 1.0‡
Vest	(%BW)	37.6 ± 5.6‡	NS	NS	42.1 ± 5.8‡
θ_t	(%BW/deg)	1.6 ± 0.6*	1.9 ± 0.3‡	0.89 ± 0.13‡	2.4 ± 0.7‡
$\Delta\theta_k$ x θ_t	(%BW/deg²)	-.166 ± .018‡	-.038 ± .008‡	.012 ± .004†	-.166 ± .018‡
Vest x $\Delta\theta_k$	(%BW/deg)	1.8 ± 0.3‡	0.67 ± 0.15‡	NS	2.0 ± 0.4‡
Vest x θ_t	(%BW/deg)	NS	NS	NS	NS
R^2		0.94	0.94	0.88	0.94

* $p < 0.05$; † $p < 0.01$; ‡ $p < 0.001$; NS = Not significant ($p \geq 0.05$).

BW = body weight;
$\Delta\theta_k$ = Peak difference in knee flexion from 70º (= peak knee flexion angle minus 70º);
Vest = Weighted vest not used (0) or used (1);
θ_t = Peak trunk flexion from vertical;
A x *B* indicates an interaction effect between variables *A* and *B*;
R^2 = Coefficient of determination between forces and those predicted by the regression model.

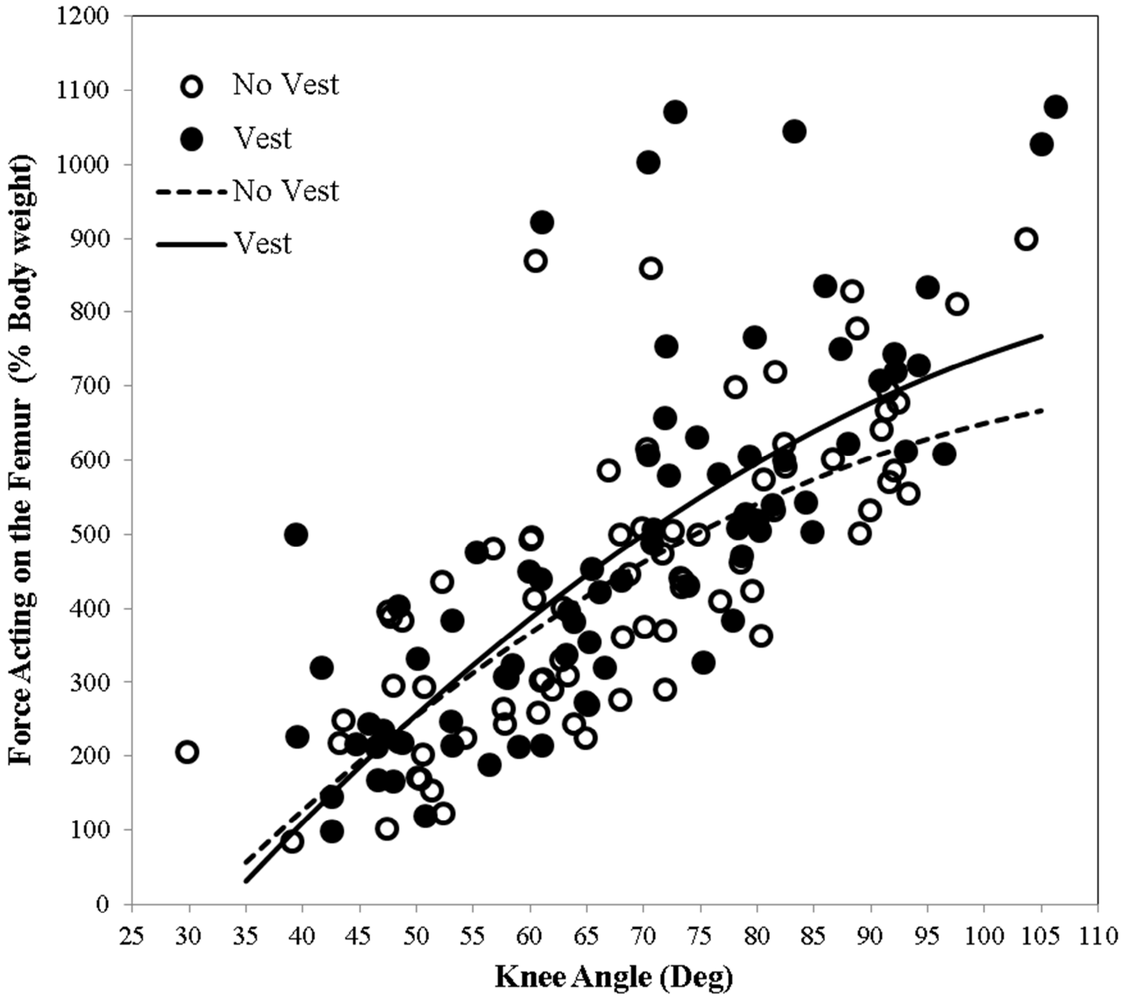

Figure 4.2: Peak distal force acting on the femur at the hip as a function of peak knee flexion angle for squats performed without and with a 5.4 kg weighted vest. Data are shown for all participants, with values averaged across the trials at a given squat depth. Curved lines represent predicted values of force based on the derived regression model (Table 4.1), with peak trunk flexion estimated from peak knee flexion using linear regression across all trials of all participants. The regression equation between peak knee angle (θ_k) and peak trunk angle (θ_t) was $\boldsymbol{\theta_t = \theta_k * 0.49 - 8.80^{\circ}}$.

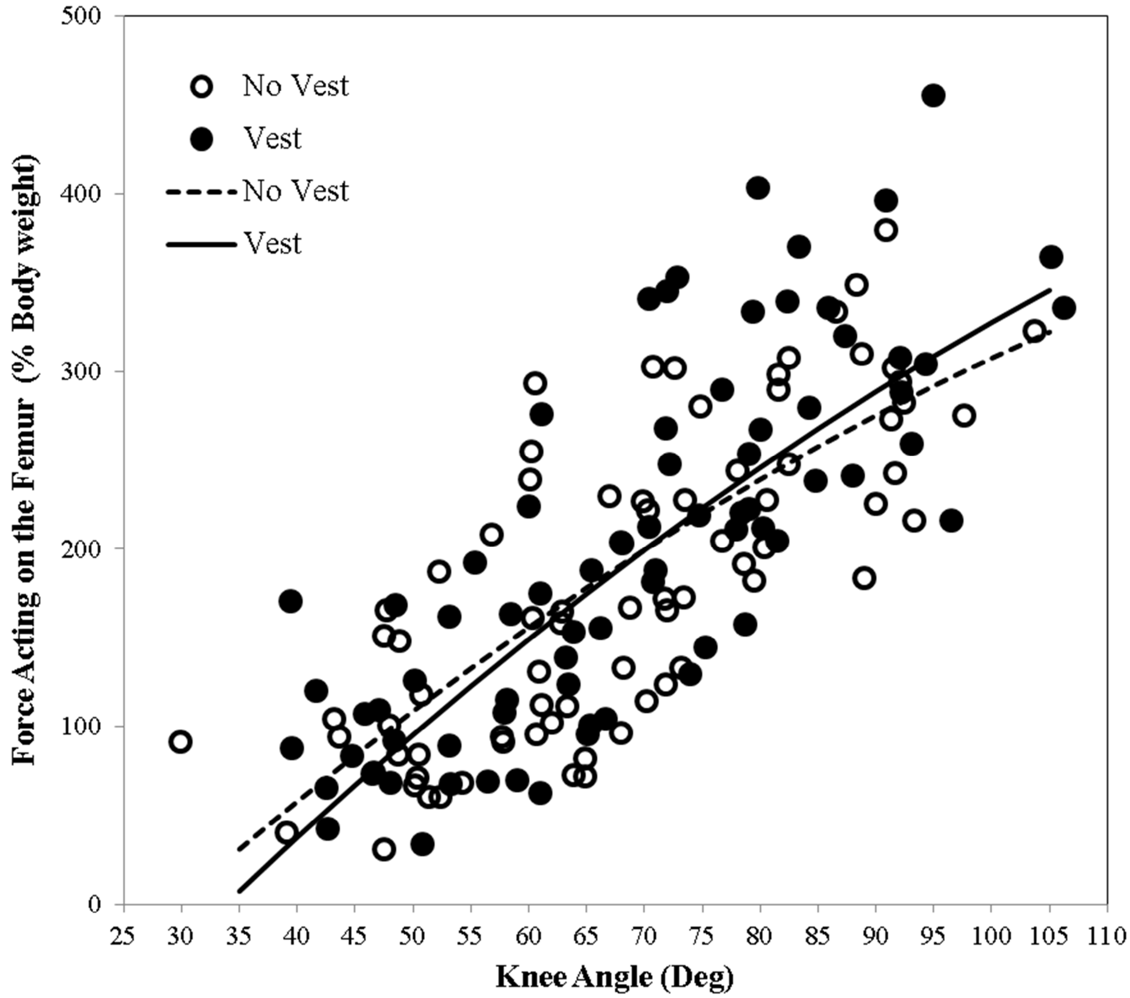

Figure 4.3: Peak lateral force acting on the femur at the hip as a function of peak knee flexion angle for squats performed without and with a 5.4 kg weighted vest. Data are shown for all participants, with values averaged across the trials at a given squat depth. Curved lines represent predicted values of force based on the derived regression model (Table 4.1), with calculations made in a similar manner as in Figure 4.2.

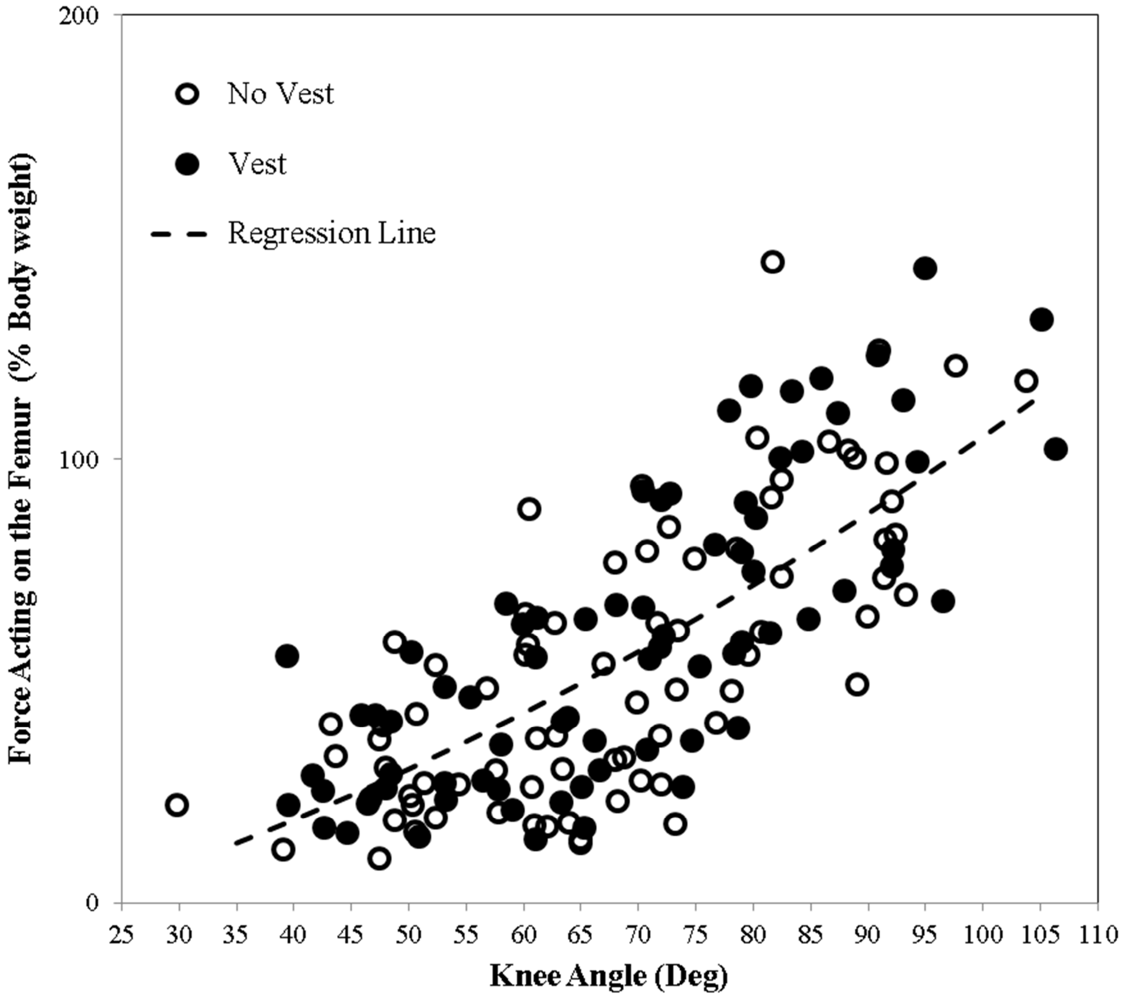

Figure 4.4: Peak posterior force acting on the femur at the hip as a function of peak knee flexion angle for squats performed without and with a 5.4 kg weighted vest. Data are shown for all participants, with values averaged across the trials at a given squat depth. The curved, dashed line represents the predicted values of force based on the derived regression model (Table 4.1), with calculations made in a similar manner as in Figure 4.2.

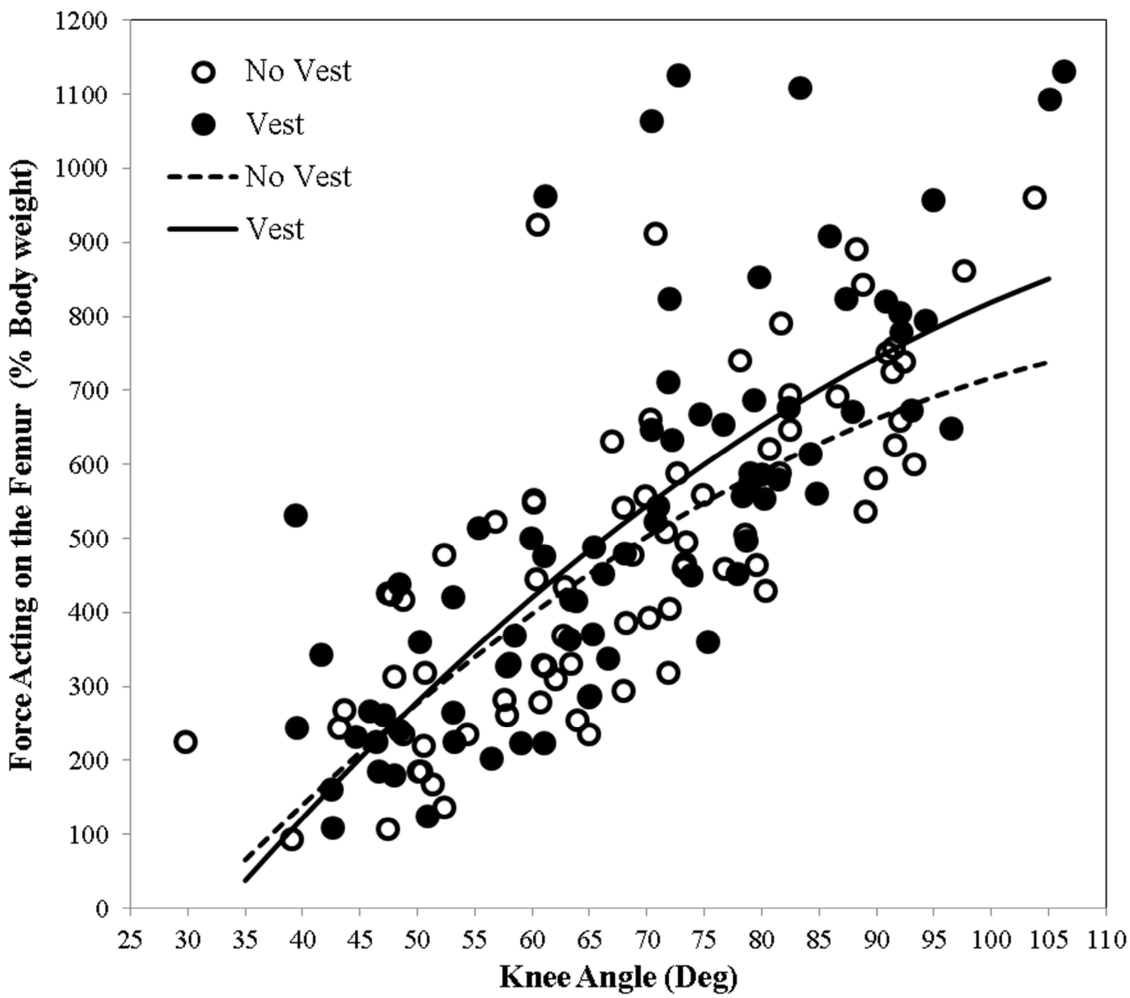

Figure 4.5: Peak magnitude of the force acting on the femur at the hip as a function of peak knee flexion angle for squats performed without and with a 5.4 kg weighted vest. Data are shown for all participants, with values averaged across the trials at a given squat depth. Curved lines represent predicted values of force based on the derived regression model (Table 4.1), with calculations made in a similar manner as in Figure 4.2.

CHAPTER 5 - DISCUSSION

As women get older, bone loss due to decreased estrogen levels can increase the risk of hip fractures (Lane, 2006). This is of concern in that hip fractures are the most severe (Johnell and Kanis, 2005) and costly (Burge et al., 2007) form of osteoporotic fracture. However, by using exercise in the form of resistance training, it may be possible for women to prevent bone loss and the associated increase in hip fracture risk. Exercise that includes resistance training can produce increases in bone mineral density (BMD) by loading bone in a way that will cause an osteogenic effect (Borer, 2005; Martyn-St James and Carroll, 2010). One exercise common to many intervention programs that involved lower-body resistance training and that were found to have beneficial effects on BMD at the hip is the squat exercise (Going et al., 2003; Winters-Stone and Snow, 2006). The squat exercise has been found to produce a greater magnitude of loading on the hip compared to many other lower body resistance exercises (Anderson et al., 1996). This suggests that performing squats can play a large role in maintaining or increasing hip BMD, as larger loads on bone produce a greater osteogenic response (Rubin and Lanyon, 1985). However, different approaches to the squat exercise, such as the squat depth and the addition of static resistance, had not yet been examined in regards to the loading at the hip.

It is important to determine the influences of squat depth and the addition of static resistance on hip loading during the squat exercise in order to develop the most effective resistance training programs for the prevention of hip fractures. Therefore, the objective of this study was to examine how the loading on the hip during a squat exercise varies as a function of the depth of the squat, as well as the degree to which the addition of static resistance to the upper body affects loading on the hip at different depths during a squat exercise. The central hypothesis was that loading on the hip would increase as the depth of the squat increased, and that the addition of static resistance would produce greater loading on the hip across a range of depths. Motion capture and ground reaction force data were collected on 20 women as they performed two sets of squats at four different depths: shallow, medium-shallow, medium-deep, and deep. One set of squats was done with body weight only and the other was done with added upper-body static resistance in the form of a 5.4 kg weighted vest. From the collected data, a biomechanical model estimated the force acting on the femur at the hip during each trial. Relationships of peak loading in three directions (i.e. distal, lateral, posterior) and of peak overall magnitude of loading to squat depth, as quantified by the peak knee flexion angle, were examined as a function of use or non-use of a weighted vest. There was a significant effect of peak knee flexion on all directional peak loads relative to the femur and on the peak magnitude of loading. In addition, use of the weighted vest was found to have a significant effect on the relationship between peak knee flexion and peak hip loading, except in the posterior direction relative to the femur.

Hip Loading as a Function of Squat Depth

The first specific aim was to determine the loading on the hip as a function of squat depth during a squat exercise. It was hypothesized that the loading on the hip would increase as the depth of the squat increased. Consistent with this hypothesis, it was found that there exists a relationship between squat depth and hip loading such that loading increased as the depth of the squat increased. Peak femoral loading at the hip during squats without a weighted vest increased as peak knee flexion increased, on average, by

8.1, 3.3, and 0.99% BW/deg in the distal, lateral, and posterior directions, respectively, over the range of 0-50° of peak trunk flexion (i.e. as computed for a peak trunk angle of 25°). For overall magnitude, peak loading at the hip without a weighted vest also increased as peak knee flexion increased, on average, by 8.6% BW/deg. All of these effects of peak knee flexion on peak loading were statistically significant.

In order to explain why the relationships exist between squat depth and hip loading, it is important to understand what forces are responsible for the observed loading. There are two main forces that contribute to the joint contact force at the hip: forces from the muscles acting across the joint, and the weight of the upper body. The weight of the upper body will increase the force acting distally along the femur during standing and shallower squats and will increase the force acting in the posterior direction relative to the femur during deeper depth squats. In a standing position or during a shallow squat, the thigh is positioned so that the weight of the upper body is acting on the top of the femoral head. Thus, in these positions, the weight of the upper body is creating a distal force acting on the femur. However, as the depth of the squat increases, the femur will move progressively to a position that is parallel with the ground. As the femur moves towards this position, the force created by the weight of the upper body on the femur will transfer from the distal direction to the posterior direction relative to the femur. Thus, as the squat depth increases, so would the hip contact force in the posterior direction. However, the weight of the upper body will have minimal effect on the lateral contact force at any squat depth. The weight of the upper body is a vertically-directed force, and during a squat, the thigh is never in a position such that a meaningful portion of the weight of the upper body would act laterally on the femoral head. As such, the weight of the upper body helps to explain the relationship between squat depth and peak loading only in the posterior direction, and only in part. The most probable explanation for the observed relationships between peak loading and squat depth across directions is that the force created by the hip extensors increases as the depth of the squat increases. This relationship is largely a result of the lower body's center of mass moving further backward as the depth of the squat increases. In order to maintain static equilibrium during this action, the trunk must flex forward so the body's center of mass is maintained over the base of support and does not move posterior to the heels. As trunk flexion increases, the center of mass of the upper body is moved more forward relative to the hip, effectively increasing the moment arm of the weight of the upper body about the hip, thereby increasing the moment created by the weight about the hip. The requirement for static equilibrium dictates that the internal hip extension moment must balance out the external moment created by the weight of the upper body about the hip. Thus, as the moment created by the weight of the upper body about the hip increases, the internal hip extension moment must also increase. Increases in the internal hip extension moment are most likely produced through increases in muscle force resulting from increased activation of the hip extensors (Caterisano et al., 2002; Wretenberg et al., 1993). As peak trunk flexion was correlated to peak knee flexion ($r = 0.66$), deeper-depth squats were associated with greater trunk flexion. Thus, increases in trunk flexion at deeper squat depths could lead to increases in the forces acting across the hip from the hip extensors, leading in turn to the observed increases in the loading of the femur at the hip.

It is also possible for the depth of a squat to influence hip loading even if the trunk angle remains unchanged. If the trunk angle remained constant during a squat, then ankle dorsiflexion would be used instead of trunk flexion to maintain static balance. In this case, the required internal hip extension moment would not change with squat depth; however, as the depth of the squat increased, so would flexion at the hip. Increased hip flexion would consequently decrease the moment arm of the hip extensors (Németh and Ohlsén, 1985), which would necessitate a greater muscle force to maintain the same hip extension moment. The resulting greater muscle forces acting across the hip could lead to increases in applied loads at the hip.

As mentioned earlier, the peak trunk flexion during the squat exercise can possibly affect hip loading by influencing the internal hip extension moment required for equilibrium. The results of the study display a significant effect of peak trunk flexion on peak hip loading across all three loading directions relative to the femur. Increases in peak trunk flexion increased femoral loading at the hip by 1.6, 1.9, and 0.89% BW/deg in the distal, lateral, and posterior directions, respectively, on average over the range of 30-110° of peak knee flexion (i.e. as determined for a peak knee angle of 70°). Increases in peak trunk flexion also increased the peak overall magnitude of the loading at the hip by 2.4% BW/deg, on average. This result is logical. As described earlier, increases in trunk flexion lead to larger required internal hip extension moments, which can lead to increases in hip muscle activity and resulting increases in hip loading.

Although both peak knee flexion and peak trunk flexion were found to affect peak hip loading, these effects were not independent of one another. There was an interaction between the effects of peak knee flexion and peak trunk flexion on peak femoral loading at the hip for all three loading directions relative to the femur and for the peak overall magnitude of loading. In the posterior direction, this interaction had a direct relationship.

That is, the effect of increased knee flexion on peak hip loading in the posterior direction increased with increasing trunk flexion. The knee flexion by trunk flexion interaction had an inverse relationship in the distal and lateral directions, as well as on the overall magnitude of loading. Consequently, the effect of increased knee flexion on peak femoral loading in the distal and lateral directions at the hip, as well as on the overall magnitude of loading, decreased with increasing trunk flexion. These results imply that, in all directions except for the posterior direction, an increased trunk flexion angle during squats will increase peak femoral loading at the hip but decrease the effect of changes in peak knee flexion on loading. An explanation for this result is based on the mechanics of the squat. In order to maintain static equilibrium, the internal hip extension moment needed can be calculated using the following equation: (moment arm of the weight of the upper body) * (weight of the upper body) * (sine of the trunk angle). Due to the nature of the sine function from 0° to 90°, which is to have progressively smaller increases in value with successive changes in angles moving toward 90°, increases in trunk flexion at lower trunk flexion angles will have a greater effect on the required hip extension moment than will increases in trunk flexion at higher trunk flexion angles. As noted earlier, the hip extension moment may be directly related to the activity of hip muscles (Caterisano et al., 2002; Wretenberg et al., 1993), which in turn may influence loading at the hip. Thus, hip loading may be more sensitive to changes in trunk flexion at lower trunk flexion angles and less sensitive to changes in trunk flexion at higher trunk flexion angles. The requirements for static balance also dictate that peak knee flexion and peak trunk flexion during a squat will be directly related to one another, as was seen in the present results. If hip loading is less sensitive to changes in trunk flexion at larger trunk flexion angles and greater trunk flexion is accompanied by greater knee flexion, then it is reasonable that hip loading would also be less sensitive to changes in trunk flexion at larger knee flexion angles. It will be noted that this is equivalent to viewing the interaction of knee flexion and trunk flexion as first mentioned, with increasing trunk flexion decreasing the effect of increased knee flexion on hip loading.

In the posterior direction, the effect of increased knee flexion on hip loading was also dependent upon trunk flexion. As trunk flexion increased, the effect of knee flexion on hip loading also increased. This relationship may be due to the position of the thigh during deeper-depth squats. Squats with higher knee flexion may influence the direction of the muscle forces acting across the hip joint, as the position of the femur is constantly changing. It is possible that the anteroposterior component of the net muscle force acting across the hip is more sensitive to changes in hip flexion at large trunk and/or knee angles, as hip muscles may pull in slightly different directions based upon the orientation of the thigh. Thus, due to the position of the femur at deeper squat depths, an increase in trunk flexion may have a greater effect on the effect of knee flexion on hip loading in the posterior direction at higher knee flexion angles.

The observed relationship between squat depth and hip loading suggests that squatting to greater depths produces higher loading on the hip. However, the observed relationship between trunk angle and hip loading suggests that increased trunk flexion during a squat can decrease the effect of knee flexion on hip loading. Because of this relationship, the regression equation theoretically suggests that, at large trunk flexion angles (> 76.5°), an increase in squat depth can actually decrease the overall magnitude of hip loading. Although the equation suggests that this happens, it is most likely a result of modeling a nonlinear effect using linear regression. Thus, even though increased trunk flexion decreases the effect of knee flexion on hip loading, it is likely that, even at high trunk flexion angles, there is still an increase in loading with an increase in knee flexion. The increase in loading due to the increase in knee flexion will just become smaller as trunk flexion increases. Similarly, the results suggest that an increase in trunk flexion will act to increase the magnitude of hip loading, even at deep squat depths (i.e. high knee flexion angles); the increase in loading due to the increase in trunk flexion will just become smaller as knee flexion increases. When the results regarding the effects of peak knee flexion and peak trunk flexion are considered together, it would appear that, in practical terms, greater loads at the hip are observed with squats at deeper depths and/or higher trunk flexion angles (Figure 4.5). Therefore, achieving higher peak knee flexion and peak trunk flexion at greater depths can be the most effective means of maximizing loading at the hip during the squat exercise.

Effect of Added Static Resistance

The second specific aim was to determine the extent to which the addition of static resistance to the upper body affects the relationship between squat depth and hip loading during a squat exercise. It was hypothesized that, across a range of squat depths, performing a squat with added upper-body static resistance would produce greater loading on the hip than performing a squat to the same depth without upper-body static resistance. The hypothesis was partially supported, as it was found that adding upper-body static resistance increased the peak distal and lateral loading of the femur at the hip at deeper squat depths only, whereas it had no effect on peak loading in the posterior direction relative to the femur. The derived regression models indicated that, in the reference condition (i.e. 70° of knee flexion, 0° of trunk flexion, no weighted vest), adding the weighted vest would result in no significant change in femoral loading at the hip in the lateral and posterior directions, but would significantly increase the peak distal and overall loads by 37.6 and 42.1% BW, respectively. The weighted vest by knee flexion angle interaction was significant in the distal and lateral directions, as well as on the overall magnitude of loading. However, this interaction was not significant in the posterior direction. The weighted vest increased the effect of knee flexion on peak femoral loading by 1.8 and 0.67% BW/deg in the distal and lateral directions, respectively, and increased the effect of knee flexion on the peak overall magnitude of loading by 2.0% BW/deg. The net result is that squats with static upper-body resistance can lead to higher observed loads at the hip compared to squats without static resistance. However, it must be noted that, according to the regression equation for the overall magnitude of hip loading, the weighted vest had minimal effect (< 12% BW) on hip loading at smaller knee flexion angles (< 55°) and a greater effect at larger knee flexion angles (Figure 4.5).

There is a mechanical reason for observed increases in hip loading through use of the weighted vest. As mentioned previously, the weight of the upper body and the muscle forces acting across the joint are the two contributors to the joint contact force at the hip. The addition of a 5.4 kg weighted vest to the shoulders will effectively increase the upper body's weight and raise its center of mass. The raising of the center of mass will increase the moment arm of the external load acting about the hip. Regarding the weight of the upper body as a contributor to hip contact force, it is logical that an increase in upper-body weight will increase the joint contact force at the hip in the distal direction relative to the femur at lower knee flexion angles and in the posterior direction relative to the femur at higher knee flexion angles. With regards to the muscle forces acting across the joint, as the weight of the upper body and its moment arm about the hip increase, the external moment created by the weight of the upper body about the hip will also increase. As the moment created by the upper body about the hip increases, the internal hip extension moment needed increases, which is created by greater muscle activation (Caterisano et al., 2002), which in turn, can lead to greater loading at the hip. However, at a given knee flexion angle, there was no observed effect of the vest on femoral loading in the posterior direction. Qualitatively, the vest appears to have had a smaller effect on loading in the posterior direction compared to the other directions (Figure 4.4), resulting in a smaller effect size. Thus, it is possible that there was insufficient power to detect an increase in hip loading by the weighted vest in the posterior direction. However, it is also probable that any such small increase would be clinically meaningless.

As effects of the weighted vest on hip loading were only observed at higher knee flexion angles (Figure 4.5), it appears that there are other aspects that are influencing the effect of the addition of static resistance on hip loading. One possible aspect is the peak trunk flexion during the squat, as higher peak trunk flexion was associated with higher peak knee flexion. As was noted, the weighted vest raises the upper body’s center of mass and effectively increases the weight of the upper body, both of which will increase the external moment about the hip created by the weight of the upper body. This moment is directly related to trunk flexion, specifically the sine of the trunk angle. Hence, the addition of the vest should produce progressively greater increases in the external moment about the hip at greater trunk flexion angles, resulting in greater increases in the internal hip extension moments needed, and thus greater increases in hip extensor activation and in force at the hip. Thus, despite the fact that the interaction effect between weighted vest use and peak trunk angle was not significant for any directional loading, the fact that effects of the weighted vest were observed only at deeper squat depths may reflect an indirect effect of trunk flexion on the effect of the weighted vest on hip loading.

The observed relationship between added static resistance and hip loading during the squat exercise would imply that, to obtain a given load on the hip, a woman can squat to a shallower depth when wearing a weighted vest than when without a weighted vest. However, in order to receive this benefit, the desired load must require a squat depth that is greater than 55°. For example, based upon the derived regression equation, a woman who desires to reach an overall loading magnitude of 600% BW at the hip would have to squat to a depth of 81.6° of knee flexion without a vest and a depth of only 75.4° of knee flexion with a 5.4 kg weighted vest, assuming 25° of trunk flexion, which was the average across all trials. This predicted reduction in required knee flexion would be greater for trunk flexion angles greater than 25° and smaller for trunk flexion angles less than 25°. This again suggests that there is an indirect effect of trunk angle on the effect of weighted vest, as discussed previously. It will also be noted that the predicted reduction in required knee flexion associated with addition of the vest is higher when the desired hip load is greater than 600% BW, and lower if the desired load is less than 600% BW (Figure 4.5). Additionally, this relationship only applies to the vest used in this study, which had a mass of 5.4 kg. Weighted vests that have a mass greater or less than 5.4 kg may have a different magnitude of effect on hip loading as a function of squat depth. Weighted vests that have a greater mass may further reduce the required knee flexion for a desired hip load, whereas vests that have less mass may reduce to lesser extent the required knee flexion for a desired load. Thus, the reduction in the required knee angle to reach a desired hip load using a weighted vest may depend upon both the desired load and the mass of the vest.

Verification of Predicted Loads

In order to verify that the results from the AnyBody model are reasonable, it is important to compare the predicted loads of the model to other studies that used instrumentation to directly measure loads on the hip. Unfortunately, there are no studies that directly measured hip loading during the squat exercise. However, one study did measure hip loading using instrumented hip implants while participants performed a knee bend activity, which appears to have been similar to the shallow squats in this study (Bergmann et al., 2001). Additionally, hip loading was measured for sitting down and standing up from a chair, both of which are similar actions to squats (Bergmann et al., 2001). The range of resulting peak loads for the knee bend activity was 117-177% BW. The resulting peak loads for sitting down and standing up were, on average, 156 and 190% BW, respectively. These loads are considerably less than most of the peak overall loading magnitudes computed in the current study (≥ 300% BW past 52° of knee flexion). Regarding the knee bending activity, it was not clear how much knee flexion was used during that activity. However, there were several shallow squats in this study that had peak overall loads ranging from 100-200% BW (Figure 4.5), which is comparable to the range of peak loads observed during the knee bend activity. Regarding the observed loads during sitting down and standing up, individuals use the chair to help break their downward movement when sitting down into a chair, and use momentum at seat-off to help propel themselves upward when standing up from a chair (Kralj et al., 1990). In contrast, during a squat, individuals are responsible for generating all of the forces needed to break the body’s downward motion and then propel it upward, as there is no chair to aid them. This difference between a squat and getting in/out of a chair may possibly lead to larger loads being placed on the hip during squats. It is also possible to use inverse dynamics to provide face validity for the loads computed in this study. Based on the average peak ground reaction forces of 78% BW across deep squat trials, an average participant thigh length of 39.6 cm, and an estimated average moment arm of 4.5 cm of the hip extensors (Németh and Ohlsén, 1985), it was estimated that the loading on the hip would be approximately 600% BW at 90° of knee flexion. This resultant loading was well within the range of observed loads during deep squats of most participants. Taking all of the previous information into consideration, it may be concluded that the hip loading computed by the AnyBody model is reasonable.

Implications for the Prevention of Hip Fractures

Bone is a highly adaptive tissue that responds to repeated higher-than-normal mechanical loading (Wolff, 1892). Bone will respond differently to differing amounts and types of loading, as dynamic and high-magnitude loads have been shown to yield the greatest amount of increase in BMD (Lanyon and Rubin, 1984; Rubin and Lanyon, 1985). Bone also responds based upon the location of applied loads, as it is possible to increase BMD at specific sites in the body by using specific exercises that target muscle groups at such sites (Winters-Stone and Snow, 2006). Several studies have determined that exercise programs using squats can produce smaller reductions in, maintain, and/or increase BMD at the hip (Going et al., 2003; Kemmler et al., 2012; Korpelainen et al., 2006; Winters-Stone and Snow, 2006). However, it is difficult to determine the exact effect of squats on BMD, as each study used a multitude of different exercises in conjunction with squats. This study attempted to provide a better understanding of the influences of the squat exercise on hip loading. This was done by quantifying the relationship between squat depth and loading on the hip, and by examining the effect of added static resistance. This relationship can give practitioners information about different approaches to the squat exercise that may result in higher loads at the hip. Consequently, higher loading at the hip may lead to an increase in BMD, which in turn may lead to a reduction in the risk of hip fractures.

It is important to determine whether the observed loads on the femoral neck in the current study will have an effect on hip BMD. Unfortunately, there is no literature that explicitly quantifies an osteogenic threshold of loading on the human femoral neck. However, such a threshold might be estimated. In rat tibiae, loading that produced strains exceeding 1050 µstrain significantly increased bone formation, whereas strains below 1050 µstrain had no evidence of increased bone formation (Turner et al., 1994). The force needed to produce a 1050 µstrain in the femoral neck can be estimated using the approximation of a cantilevered hollow tube with a cortical bone stiffness of 17 GPa (Nordin and Frankel, 2001), a total cross-sectional area of 749.4 mm^2 (Manske et al., 2006), a cortical area of 132.5 mm^2 (Manske et al., 2006), and a femoral neck length of 58 mm (Michelotti and Clark, 1999). It must be noted that different directional loading relative to the femur will create different types of stress in the femoral neck. Loading of the femur in the distal direction at the hip creates a bending moment that applies tensile stress to the superior aspect of the femoral neck and compressive stress to the inferior aspect. Similarly, loading in the posterior direction creates a bending moment that applies tensile stress to the anterior aspect of the femoral neck and compressive stress to the posterior aspect. Lateral loading will create compressive stress in all areas along the femoral neck. Due to the principle of superposition, loading in all three directions will combine to create different resulting stresses on different areas of the femoral neck. The corresponding areas can be divided into four quadrants: inferior, superior, posterior, and anterior. At a point midway along the neck of the femur, women would have to squat to depths of approximately 46º of knee flexion to exceed 1050 µstrain throughout the cortical regions of the inferior and superior quadrants of the femoral neck (Figure 5.1). This is regardless of whether or not the weighted vest was used. The majority of this strain will result from the bending produced by the distal force. However, at a point midway along the neck of the femur, it would be impossible for women to exceed 1050 µstrain throughout the cortical regions of the anterior and posterior quadrants of the femoral neck, even if the weighted vest was used (Figure 5.1). This is not to say that the majority of the cortical bone in the posterior and anterior quadrants of the femoral neck would not exceed 1050 µstrain. At 70° of knee flexion and 25° of trunk flexion without a weighted vest, the superimposed strain would predict that approximately 86% of the inner circumference would be above the estimated strain threshold. The use of the weighted vest would only increase this number to 87%. If knee flexion was increased to 90° and trunk flexion increased to 36°, approximately 89% of the inner circumference would be above the estimated threshold without a weighted vest, and 90% with the weighted vest. Of note though, it would be impossible for the entire inner circumference to reach the estimated strain threshold during a squat, regardless of the squat depth or amount of static resistance used. Nevertheless, based upon the above calculations, it would appear that the majority of the loads experienced in this study would actually excite some sort of osteogenic response in some areas of the femoral neck, but not others. However, the exact magnitude of this response is unknown and could be investigated in future research.

The results of this study give practitioners the suggestion that squatting to a deeper depth can be more beneficial for bone health at the hip than squatting to a shallower depth. However, practitioners must also consider the angle of the trunk during squats. Women can be encouraged to use as much trunk flexion as they feel comfortable with during a squat, as it was shown that increased trunk flexion was associated with increased hip loading. The results of this study also suggest that adding upper body static resistance in the form of a weighted vest allows women to squat to shallower depths while still obtaining the high loading benefit of deeper-depth squats performed without static resistance. However, this finding can only be applied when squatting to depths past 55° of knee flexion. Thus, practitioners who encounter women who have trouble squatting to deep depths can either prescribe increased trunk flexion during shallow and/or medium-shallow squats, or prescribe the use of a weighted vest if the woman's peak knee flexion is greater than 55°. However, in order to prevent excessive stress on the musculature at the hip and lower back, increases in trunk flexion or the addition of a weighted vest during squats should be introduced gradually into an exercise intervention.

In conclusion, for the current population of middle-aged women, the results of this study suggest that women should squat to depths greater than 46° of peak knee flexion in order to possibly produce an osteogenic effect throughout the cortical regions of the inferior and superior quadrants of the femoral neck. Additionally, the results indicate that, for peak knee flexion angles greater than 55°, using static resistance in the form of a 5.4 kg weighted vest can further increase the overall magnitude of loading on the hip during the squat exercise. Therefore the weighted vest should have some clinical applications to squatting at deeper depths. For such squats, the effect of the weighted vest on hip loading could allow women to use a weighted vest in order to obtain either the same benefits for bone health at the hip at lesser squat depths or greater benefits at the same squat depth, compared to squats without a weighted vest. More research needs to be done to determine how greater increases in resistance, different exercises, and age affect loading on the hip. Such information will give practitioners further insight that may help them to identify different approaches to exercise that can be effective in an exercise-based hip fracture prevention program.

Assumptions and Limitations

Several biomechanical models exist to determine loads acting on the body, all of which have their limitations. An accurate model to assess loading on the hip should include both body position and hip muscle activity (Németh et al., 1984). In keeping with this, the AnyBody Modeling System (AnyBody Technology, Aalborg, Denmark) used in this study uses motion optimization to determine body segment movements and inverse dynamics to determine muscle forces in the lower limb. However, the AnyBody model is a quasi-static model, which does not take into account the dynamics of muscle activation. This characteristic can create possible jumps in estimated muscle force, which will result in errors in the estimated joint contact force at the hip, both of which can be major limitations of the study. However, the squat exercise is a relatively slow-motion task that does not require rapid changes in muscle activation, making the adverse effect of neglecting muscle activation dynamics on the results negligible (Damsgaard et al., 2006). Another limitation of the AnyBody model is in its scaling of inertial properties and strength to each woman based on a variety of scaling functions using segment lengths and masses, as well as estimated percentage of body fat. Considering that the AnyBody model used was based on the anatomy of a man, and that every individual is geometrically different, the scaling of the model will better fit some individuals than others. As such, the scaling could introduce errors by under- or overestimating segment masses and strength, which could adversely influence the estimated muscular activity levels and joint contact forces. Errors in scaling could also lead to errors in muscle geometry, affecting the moment arms of the muscles. Increases in the moment arms of muscles will effectively decrease the force generated by the muscle in order to produce the required torque. The opposite is true for decreases in the moment arms of muscles. In turn, errors in the estimated muscle forces could lead to errors in the calculation of joint contact forces. An additional limitation is the cost function used to optimize the muscle recruitment pattern, which was minimizing the sum of the cubed muscle activations. Muscles can be recruited in many different patterns, and assuming the muscles follow a particular recruitment pattern will create a level of error within the results. The final limitation of the AnyBody model is that it was typically not able to generate the muscle forces required to duplicate the recorded motion without exceeding its strength capabilities, and it repeatedly placed the activity of selected muscles over 100%. The effect of this on the results is unknown. However, based on previous studies that have compared the AnyBody model output to literature that used direct measurement (Wehner et al., 2009; Wehner et al., 2010), any error that is an outcome of the model should be relatively small from trial to trial, rendering it an acceptable limitation.

Due to the biomechanical model used in the analysis, underestimation of muscle coactivation is another limitation of this study. Underestimation of muscle coactivation can result in an underestimation of muscle activity and joint contact forces. However, as was noted in the review of literature, when examining muscle activity during squats at 40, 60, and 90 degrees of knee flexion with both a flexed and a straight trunk, it was found that coactivation of the rectus femoris and biceps femoris was present only when individuals squatted to 40 degrees of knee flexion with a flexed trunk (Sousa et al., 2007). Considering that most of the trials in this study were performed with a knee flexion angle greater than 40 degrees, it is reasonable to assume that underestimation of muscle coactivation did not introduce large amounts of error into the computed forces at the hip. However, in order to produce the most accurate results, electromyography (EMG) measures might have been used to determine levels of muscular activity.

Another limitation of this study is that it only investigated loading on the femoral neck. Because intertrochanteric fractures are equally likely as fractures to the femoral neck (Gallagher et al., 1980), it would have been beneficial to also examine loading on the greater trochanter of the femur. Additionally, there are many different hip muscles activated when performing the squat exercise. It would have also been beneficial to determine the specific muscles that contribute the largest amount of hip loading, as this could provide further information on other exercises involving similar muscle groups that might create high loads at the hip. An analysis of the different loadings on specific areas of the hip, as well as of the activity levels of specific hip muscles, could be a follow-up to this study.

Variations in the mechanics of the squat exercise could also be a limitation to this study. Due to the wide age-range of the sample population, it is almost certain there would be some variations in squatting techniques between women, specifically in trunk flexion and squat speed. The only way to address these variations was to provide thorough instructions to the participants that gave them a model on how to perform a squat with the desired technique. Trunk flexion and squat speed were somewhat controlled by instructing participants not to bring their shoulders past the tips of their toes while squatting and through the use of a metronome in the warm-up, respectively. Although participants were able to proficiently follow the guidelines regarding their trunk position and to maintain a consistent speed across all squat depths, there was undoubtedly some variability between depths and trials. Differences in fitness level and in experience with performing squats most likely also contributed to differences in squat technique between participants. However, different squatting techniques may, in fact, enhance external validity, which would allow the results to apply to a larger population. This enhancement of external validity renders variations in squat technique to be an acceptable limitation.

Another limitation of the study is the fact that it only used one mass of vest, which was 5.4 kg. It would have been beneficial to possibly use two weighted vests of different masses in order to further quantify differences in the effect of static resistance on hip loading during squats. Additionally, the weighted vest was not scaled to a percentage of body mass. The vest used represented a percentage of body mass that ranged from 6.3-10.8% across participants, with an average of 8.4%. The fixed mass of the vest could have had a greater effect on hip loading among participants of lower mass compared to participants with higher mass. However, in order to get every participant to the average of 8.4% added to her body mass, the median and 75^{th} percentile amounts of mass that would have had to have been added or subtracted from the weighted vest would have been 0.4 kg and 0.8 kg, respectively. As such, the difference in the effect of the weighted vest on loading between participants was most likely negligible.

Finally, the results of this study can only be applied to the population studied and task assessed: women aged 35-50 years performing different-depth squats with and without added upper-body resistance. The results cannot be extended to other lower body exercises, men, or older populations. However, these are not major limitations, as the focus of the study is on the prevention of bone loss and the associated hip fractures, both of which can have an immediate and/or near-future impact on the population studied. Other tasks involving different populations and different exercises, such as lunges or step-ups, may be investigated in further studies.

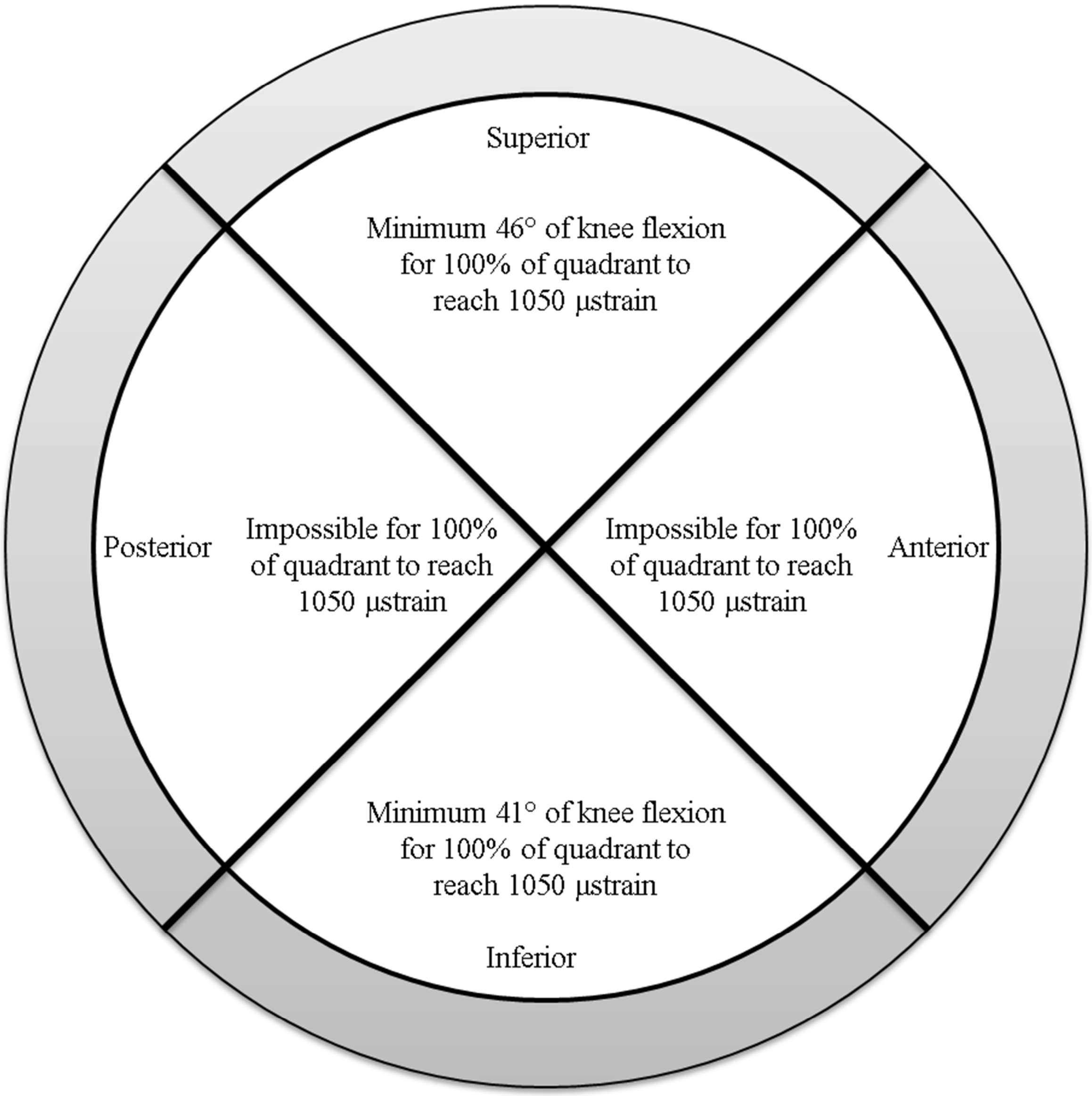

Figure 5.1: Diagram of the estimated peak knee flexion needed to create an osteogenic effect throughout the cortical bone in different quadrants of the femoral neck during a squat. The diagram is a cross-section of the femoral neck at its midpoint along the mediolateral axis. The gray area represents the cortical bone of the femoral neck, and the enclosed white area represents the cancellous bone. The peak knee flexion required to reach the ostegenic threshold of 1050 µstrain (Turner et al., 1994) throughout 100% of the cortical area is dependent upon the quadrant. Although it is impossible for the entire cortical area of the anterior and posterior quadrants to reach 1050 µstrain, greater than 50% of the cortical areas of both quadrants will be over this threshold with peak knee flexion angles greater than 50°.

CHAPTER 6 - CONCLUSION

Hip fractures are the most costly form of fracture (Burge et al., 2007) and the risk of these fractures is directly related to osteoporosis and the associated loss in bone mineral density (BMD) (Kanis, 2002). Exercise that includes resistance training can be an effective means of preventing hip fractures by preventing or slowing the loss of BMD (Engelke et al., 2006; Kemmler et al., 2012). To prevent hip fractures, it is important to incorporate exercises that train the lower body, as the response of bone density is site specific (Winters-Stone and Snow, 2006). The squat exercise has been shown to produce high loads on the hip (Anderson et al., 1996), and these high loads can lead to increases in BMD (Borer, 2005). However, different approaches to the squat exercise that would be effective in increasing loads at the hip had not been identified. In particular, effects of the depth of the squat and of added static resistance on peak loading at the hip had not been studied to date.

The objective of this study was thus to examine how the loading on the hip during a squat exercise varies as a function of the depth of the squat, as well as the degree to which the addition of static resistance to the upper body affects loading on the hip at different depths during a squat exercise. The central hypothesis was that loading on the hip would increase as squat depth increased, and that the addition of static resistance would produce greater loading on the hip across a range of depths. To test this hypothesis, two specific aims were pursued. The first specific aim was to determine the loading on the hip as a function of squat depth during a squat exercise. It was hypothesized that the loading on the hip would increase as the depth of the squat increased. The second specific aim was to determine the extent to which the addition of static resistance to the upper body affects the relationship between squat depth and hip loading during a squat exercise. It was hypothesized that, across a range of squat depths, performing a squat with added upper body static resistance would produce greater loading on the hip than performing a squat to the same depth without upper body static resistance.

Motion capture and ground reaction force data were collected from 20 women, aged 35-50 years, as they performed two sets of squats, each of which included 3-4 repetitions each of shallow, medium-shallow, medium-deep, and deep-depth squats. One set of squats was done with added upper-body static resistance in the form of a 5.4 kg weighted vest, and the other set was done without the weighted vest. After the data were collected, they were inserted into a lower-body biomechanical model that determined the force acting on the femur at the hip during each trial. Relationships of the peak loading in the distal, lateral, and posterior directions relative to the femur and of the peak overall magnitude of loading to the peak knee flexion angle were examined, as was the effect of the weighted vest on each of these peak loads.

With regards to the first specific aim, hip loading significantly increased as the depth of the squat increased, which was consistent with what was hypothesized. Significant effects of peak knee flexion, peak trunk flexion, and the knee flexion by trunk flexion interaction on hip loading were found in every direction. Increases in peak knee flexion and in peak trunk flexion were associated with increases in hip loading across all directions. However, the effect of peak knee flexion on loading depended on peak trunk flexion. Increases in trunk flexion during squatting decreased the effect of knee flexion in all directions except for the posterior direction. The combined effects of knee flexion and trunk flexion imply that, at greater peak trunk flexion angles, femoral loading in the distal and lateral directions at the hip will be less sensitive to changes in peak knee flexion. However, increases in peak knee and/or peak trunk flexion will still increase loading on the femur, regardless of the depth of the squat.

With regards to the second specific aim, the addition of upper-body static resistance increased loading on the hip only in some directions and at deeper squat depths (> 55°), which was partially consistent with what was hypothesized would be the case. In the reference condition (i.e. 70° of knee flexion, 0° of trunk flexion), a significant effect of the weighted vest on hip loading was found to increase loading in the distal direction and the overall magnitude of loading. Additionally, a significant interaction between effects of the weighted vest and knee flexion on hip loading was found in the distal and lateral directions, as well as on the overall magnitude of loading. The weighted vest increased the effect of knee flexion on hip loading in all directions except for the posterior direction.

In practical terms, for the population tested, the results of this study suggest that women can increase the peak loading on the hip during squats by squatting to a deeper depth and using upper-body static resistance in the form of a weighted vest. The results of this study give a recommendation for middle-aged women to squat to the deepest possible depth. Regarding knee flexion, women should be encouraged to reach a minimum squat depth of 46°. Based on the estimated strains produced in the femoral neck by the observed hip loading, squats beyond this depth appear capable of eliciting an osteogenic response throughout the cortical regions of the superior and inferior quadrants of the femoral neck. Practitioners should also recommend that women use as much trunk flexion as they feel comfortable with during a squat in order to experience the full benefits of the relationship between trunk flexion and hip loading. The results of this study also provide the recommendation to use a weighted vest during squats in order to increase loading on the hip, provided the depth of the squat exceeds 55° of knee flexion. If women have trouble exceeding 55° of knee flexion during a squat, practitioners can recommend increasing trunk flexion in order to increase the loads applied to the hip. This study provides groundwork for the identification of different aspects of the squat exercise that can influence hip loading and possibly aid in the prevention of hip fractures. Future research can build off this study in order to identify other recommendations to practitioners that will aid them in the implementation of exercise-based hip fracture prevention programs.

Future Research

More research should be conducted to identify different approaches to resistance exercises that influence hip loading and the prevention hip fractures. A repeat of this study with a different age range, sex, or added resistance could be effective in extending the results to a larger population. Shifting the age range to post-menopausal women or increasing the resistance of the weighted vest could possibly produce differing results, which may give practitioners more information regarding the prescription of the squat exercise.

A study that estimated the exact loading on different sites of the femur during the squat exercise using a finite element analysis would also be beneficial. This analysis would give practitioners information on how the squat exercise affects loading on different parts of the femur. Because intertrochanteric fractures are equally likely as femoral neck fractures (Gallagher et al., 1980), this information could be used to prescribe the squat exercise to individuals who are at a higher risk for such fractures. A longitudinal study that had participants perform the squat exercise based on the present recommendations would be extremely helpful in determining the specific effects of squats on hip BMD. Using a six- to twelve-month intervention involving only squats, with pre- and post-intervention measures of BMD, the exact effects of the squat exercise on bone health at the hip could be determined. Finally, a study that determined how different lower body exercises affect loading on the hip would be beneficial. Such exercises could include lunges (multidirectional) and step-ups. These further studies could use a protocol that is similar to the current study. In doing so, the influence of exercise technique and the effect of added static resistance on hip loading could be identified.

Summary

This study used a literature-validated lower-body biomechanical model to quantify the relationship between squat depth and the loading on the hip and to examine the effect of added static upper-body resistance on hip loading. It was found that loading on the hip increased with increasing peak knee flexion and peak trunk flexion. Additionally, it was found that use of a 5.4 kg weighted vest increased loading on the hip at peak knee flexion angles beyond 55°. The main conclusions that can be drawn from this study are: (1) squatting to deeper depths will produce higher loading on the hip, (2) squatting with increased trunk flexion will produce higher loading on the hip, and (3) using a 5.4 kg weighted vest will increase loading on the hip only at deeper squat depths compared to squats without using a weighted vest. It is possible that the peak loads experienced by participants in this study did, in fact, elicit an osteogenic response, yet the exact magnitude of such a response in unknown. Further research must be done in order to determine the exact osteogenic response to the squat exercise, as well as to other lower-body exercises. This research may lead to the identification of the most efficient way to achieve high loads at the hip during different exercises, which may lead to increases in BMD at the hip. The associated increases in BMD may reduce the risk of developing osteoporosis and the associated hip fractures.

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APPENDICES

Appendix A Participant Consent Form

School of Biological and Population Health Sciences
Oregon State University, College of Public Health and Human Sciences, 101 Milam Hall, Corvallis, Oregon 97331-5109
Phone 541-737-2643 | **Fax** 541-737-6914 | health.oregonstate.edu

CONSENT FORM

Project Title:	HIP LOADING DURING THE SQUAT EXERCISE
Principal Investigator:	Michael Pavol
Student Researcher:	Gabe Haberly
Co-Investigator(s):	Elizabeth Doran, Laura Lien
Version Date:	2/10/13

1. WHAT IS THE PURPOSE OF THIS FORM?

This form contains information you will need to help you decide whether to be in this study or not. Please read the form carefully and ask the study team member(s) questions about anything that is not clear.

2. WHY IS THIS STUDY BEING DONE?

As women get older, their risk of suffering a hip fracture increases due to the loss of bone that begins during menopause. One way to slow this bone loss is through exercises that apply larger-than-normal loads to the bone. The squat might be such an exercise, depending on how it is performed. The purpose of this study is therefore to examine if and how the loading at women's hips during a squat exercise is affected by the depth of the squat and by wearing a weighted vest during the exercise.

This study is being conducted for a master's thesis. The results may also be published in scientific journals, presented at scientific meetings, used to seek funding for follow-up studies, and used for educational purposes.

Up to 40 individuals may be invited to take part in this study.

3. WHY AM I BEING INVITED TO TAKE PART IN THIS STUDY?

You are being invited to take part in this study because you are a healthy adult woman, 35-50 years of age; you have regularly participated in moderate-to-high intensity physical activity for the past 4 weeks; you have never had surgery to your back, hip, or knee; you do not have osteoporosis; and you do not have a past or present injury or condition that would make participating in this study difficult or painful or that might prevent you from performing the required tasks correctly.

4. WHAT WILL HAPPEN IF I TAKE PART IN THIS RESEARCH STUDY?

If you agree to take part in this research study, you will come to the Biomechanics Laboratory at Oregon State University for testing. The study activities include completing a questionnaire, performing a warm-up, preparation for motion capture, performing sets of squat exercises, and having body measurements taken. Details are as follows:

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Oregon State University	IRB Study # 5662	Expiration Date 03/07/2014

- **Questionnaire:** You will record information about your health history and your recent physical activity on a questionnaire. We may ask you to return for testing on a different day or end your participation in this study as a result of the information you provide.
- **Warm-up:** You will change into clothes appropriate for exercise, which must include spandex shorts or other shorts of mid-thigh length or shorter. We will measure your thigh length. You will then walk for 3 minutes. Next, you will perform 3-10 deep squats and 1-3 each of shallow, medium-deep, and medium-shallow squats (described below). You will then repeat the 4 squat exercises while wearing a weighted vest. We will explain and demonstrate each exercise and will give you feedback regarding your technique.
- **Preparation for Motion Capture:** We will tape a set of 21 reflective markers to your skin and clothing. We may also tape your shirt to the back of your neck. Three more markers will be on a thin elastic strap that will be wrapped snugly around your trunk, above the waist. The motion of these markers will be recorded by our cameras during each data collection trial that follows.
- **Squat Exercises:** You will perform 2 sets of squat exercises, one while wearing a vest that weighs about 10-12 lb. and one with no added weight (that is, without the vest). Before and potentially during each set, you will be filmed one or more times while standing still. Each set of exercises will then consist of 3-4 repetitions in a row of each of the following: shallow squats, medium-shallow squats, medium-deep squats, and deep squats.

  For the squat exercises, you will stand with your feet in marked positions, arms folded across your chest, and your hands at your shoulders. A chair will be behind you. During the first part of the exercise, you will lower your hips towards the ground by flexing your hips, knees, and ankles, like sitting down onto a chair. Once your hips reach the specified squat depth, you will stand back up by extending your hips, knees, and ankles. When performing a *deep squat*, you will lower your hips as far as you feel you safely can, without touching the chair. To perform a *shallow squat*, you will lower your hips until your knees are above your toes and your hips are above your heels. For the *medium-shallow* and *medium-deep squats*, you will lower your hips until your knees are flexed about 1/3 of the way and 2/3 of the way, respectively, between a shallow squat and a deep squat. You will be asked to perform all of the squats at a speed that would let you complete a deep squat (down and back up) in about 2 seconds.

  During each repetition of the squat exercises, we will film your movements and record the forces beneath your feet. You will be given short rest periods between repetitions and at least 2 minutes of rest between sets.
- **Body Measurements:** We will measure your height, weight, foot length, ankle width, and knee width. Weight will be measured using a scale. Standing height will be measured using a type of wall-mounted ruler. The other measurements will be made using calipers.

Study duration: The testing will occur in a single session that will last about 1-1.5 hours.

Recordings: Being filmed by the motion capture system is a required part of participating in this study. You will not be identifiable in the recordings, as our cameras will only record the markers that are attached to your body. Nevertheless, you should not enroll in this study if you do not wish to be recorded.

Future contact: If you choose to provide us with your contact information, we may contact you in the future for another similar study. You may ask us to stop contacting you at any time.

5. WHAT ARE THE RISKS AND POSSIBLE DISCOMFORTS OF THIS STUDY?

The possible risks and/or discomforts associated with being in the study include: fatigue from the exercises and muscle soreness for a few days after the testing. We will need to touch you to apply and remove the reflective markers. Possible, but unlikely, risks associated with participating in the study include pulling a muscle or injuring your back, hip, or knee. If you lose your balance during a squat, you could experience a fall that results in injury, ranging in possible severity from a bruise to a broken bone. There is a risk that we could accidentally disclose information that could identify you. Finally, the security and confidentiality of information collected from you online cannot be guaranteed. Confidentiality will be kept to the extent permitted by the technology being used. Information collected online can be intercepted, corrupted, lost, destroyed, arrive late or incomplete, or contain viruses.

Several steps have been taken to reduce the risk involved in participating in this study. We have limited the number of repetitions of the squat exercises that you will perform and the weight of the vest that you will wear during one of the 2 sets of exercises. You will undergo a warm-up to prepare your muscles and we will end your participation if you find the exercises too difficult. We will also watch and correct your form during the exercises. A chair will be behind you during all squat exercises, in case you lose your balance. We will let you rest as much as you want. The exercises will be stopped if you say that you are in pain or if we judge that you cannot safely continue. You may also stop the testing at any time for any reason. In particular, **you should immediately stop exercising if you experience any pain.**

6. WHAT HAPPENS IF I AM INJURED?

Oregon State University has no program to pay for research-related injuries. If you think that you have been injured as a result of being in this study, we need you to tell us. You can do this during your testing session or by contacting Michael Pavol afterwards, either at (541) 737-5928 or at mike.pavol@oregonstate.edu. Besides telling us, you should also contact your physician.

7. WHAT ARE THE BENEFITS OF THIS STUDY?

This study is not designed to benefit you directly.

8. WILL I BE PAID FOR BEING IN THIS STUDY?

You will not be paid for being in this research study. If you complete the testing, you will receive a $5 gift card to Starbucks. If we choose to end your participation or if you choose to withdraw from the study before the testing is complete, you will not receive the gift card.

9. WILL IT COST ME ANYTHING TO BE IN THIS STUDY?

We will not reimburse you for any cost of travel to or from the campus of Oregon State University for your testing session.

10. WHO WILL SEE THE INFORMATION I GIVE?

The information you provide during this research study will be kept confidential to the extent permitted by law. Research records will be stored securely and only researchers will have access to the records. Federal 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. Some of these records could contain information that personally identifies you.

If the results of this project are published, your identity will not be made public.

To help ensure confidentiality, we will identify your data only by an assigned subject code, and not by name. In addition, the motion capture system will record only the markers that are attached to you; no identifiable images of you will be recorded or saved. Any documents that include your name will be stored in a filing cabinet in the Biomechanics Laboratory at Oregon State University. This laboratory is kept locked when not occupied by the laboratory staff.

11. WHAT OTHER CHOICES DO I HAVE IF I DO NOT TAKE PART IN THIS STUDY?

Participation in this study is voluntary. If you decide to participate, you are free to withdraw at any time without penalty. You will not be treated differently if you decide to stop taking part in the study. If you choose to withdraw from this project before it ends, the researchers may keep information collected about you and this information may be included in study reports.

Participation terminated by investigator: In some circumstances, your participation in this study may be ended without your consent. This will happen if you cannot meet the criteria for participating in the study. It will also happen if you are unable to perform the deep squats correctly during the warm-up or if you find the warm-up exercises to be painful or too difficult.

12. WHO DO I CONTACT IF I HAVE QUESTIONS?

If you have any questions about this research project, please contact: Michael Pavol, at (541) 737-5928 or by email at mike.pavol@oregonstate.edu

If you have questions about your rights or welfare as a participant, please contact the Oregon State University Institutional Review Board (IRB) Office, at (541) 737-8008 or by email at IRB@oregonstate.edu

WHAT DOES MY SIGNATURE ON THIS CONSENT FORM MEAN?

Your signature indicates that this 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.

Do not sign after the expiration date: 03/07/2014

Participant's Name (printed): ____

______________________________________ ______________________________

(Signature of Participant) (Date)

______________________________________ ______________________________

(Signature of Person Obtaining Consent) (Date)

Appendix B Physical Activity and Health History Questionnaire

OREGON STATE UNIVERSITY
BIOMECHANICS LABORATORY

Subject Code: ____________

HIP LOADING DURING THE SQUAT EXERCISE

Physical Activity and Health History Questionnaire

Personal Information:

Age: _______

Physical Activity History:

Moderate-to-high intensity physical activity includes such activities as strength training, yoga/Pilates, aerobics, dance, swimming, bicycling, running, etc.

☐ Yes ☐ No Did you participate in 20 minutes or more of moderate-to-high intensity physical activity on at least 2 days per week for each of the past 4 weeks?

Health History:

A *squat exercise* is like sitting down and then standing back up, except without a chair.

Do you have a past or present injury or condition that would make it difficult or painful for you to perform:

☐ Yes ☐ No A squat exercise

☐ Yes ☐ No A squat exercise while wearing a vest that weighs 10-12 lb.

Have you ever had any of the following?

☐ Yes ☐ No Chronic low back pain or a serious back injury

☐ Yes ☐ No Surgery on your back, hip, or knee

Have you had any of the following **in the past 6 months**?

☐ Yes ☐ No Balance problems or dizziness

☐ Yes ☐ No Back pain (not including mild soreness)

☐ Yes ☐ No Broken bone in your lower body (from the waist down)

☐ Yes ☐ No Head injury, concussion, or loss of consciousness (e.g. fainting)

☐ Yes ☐ No Pregnancy

☐ Yes ☐ No Surgery (not including dental surgery)

Do you **currently** have any of the following?

☐ Yes ☐ No Osteoporosis or bone disease

☐ Yes ☐ No Neurological problems or conditions (e.g. epilepsy, Multiple Sclerosis, Parkinson's)

☐ Yes ☐ No A heart or lung problem that limits your ability to exercise

☐ Yes ☐ No Cold, flu, or sinus symptoms

☐ Yes ☐ No Have you taken any of the following types of drugs or medications **in the past 24 hours**?

- Alcohol (2 or more beers, glasses of wine, or "hard" alcoholic drinks)
- Sedatives or anxiety/tension relief medication (e.g. Halcion, Xanax, Phenobarbital)
- Recreational drugs (e.g. marijuana, cocaine)
- Antihistamines (excluding non-drowsy, used as directed)
- Anti-inflammatory medication/pain relievers (e.g. aspirin, Ibuprofen)

Continued on other side

☐ Yes ☐ No Is there any other information that you feel we should know about your health? If Yes, please explain.