Measuring femoral neck loads in older adults during stair ascent and descent

by

Chen Deng

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Program of Study Committee:
Timothy R. Derrick, Major Professor
Jason C. Gillette
Wei Hong

Iowa State University

Ames, Iowa

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ABSTRACT

Fractures at the femoral neck play an important role in morbidity and mortality among older adults. Understanding the loading environment is a crucial factor for reducing the incidence of fracture at this site. In this study, the stresses on the femoral neck during stair ascent and descent were estimated. These femoral neck stresses were also compared to the peak loading from internal hip moments. Five male and five female adult subjects performed 5 successful trials of stair ascent and the same number trials of descent with a 3-step staircase. Motion and force data were collected. Inverse dynamics was used to calculate 3-D joint moments and reaction forces at the hip, knee and ankle joints of the right leg. Musculoskeletal model and static optimization were used and muscle forces, joint reaction forces and moments were used to estimate the 3-D moments and forces at the midpoint of the femoral neck. A standardized elliptical model of the bone structure was used to estimate the stresses of the model. Differences in peak stresses and moments were assessed by dependent t-tests ($p < .05$).

The peak hip extensor moment was significantly greater during ascent as compared to descent ($p = .001$). The 1st peak tensile stress was significantly increased at the superior site during the descent condition ($p = 0.005$), but the 2nd peak showed no significant difference between stair ascent and descent ($p = .098$). Both peak compressive stresses at the inferior site during showed no significant differences between stair ascent and descent (1st peak: $p =.105$; 2nd peak: $p = .071$). Conclusions

concerning the loading of the proximal femur were contradictory depending on if loading was assessed via hip joint moments or from femoral neck stresses. The results of this study indicate that researchers may benefit from a more comprehensive evaluation of the loading environment by estimating bone stresses as well as joint moments during stair ascent and descent.

CHAPTER 1. INTRODUCTION

Fractures are among the most serious injuries for people, but the causes can vary. Repetitive or highly loaded activities (Turner et al. 2005), the loss of BMD (McCreadie et al. 2000), and a variety of illnesses can be main causes of a fracture.

For people, especially for older ones, hip fracture can be the most common type of fracture and one of the most serious. For females, the incidence of fracture of the femoral neck doubles every 5 years after the age of 60, and the average risk at age 80 is about 7 percent (Jensen et al. 1980). Fractures at the femoral neck play an important role in morbidity and mortality among people, especially older individuals (Graves et al. 1992). High load and repetitions are crucial risk factors for fractures. To explore the mechanism of femoral neck fractures, the loads at the femoral neck in different activities (Bergmann et al. 1995), including stair ascent, descent and running, need to be known. The loads could be presented in the form of stresses, forces or moments at the hip joint or femur.

To investigate the loading at the femoral neck for different activities, we need to find out reaction forces and moments as well as the muscle forces at the hip joint. Keyak et al. (2001) worked on cadaver proximal femora using CT scan-based linear finite element models. Two types of loading were used: fall and atraumatic loading conditions. The results showed us the values and directions of minimum and maximum fracture load for both conditions. But it didn’t show us if the real directions of applied force during

daily activities were similar to their settings. This problem was also in another study from Swiontkowski et al. (1987), in which the average bending and torsion parameters were reported through a material testing of cadaver proximal femora.

Several studies directly measured hip contact force and moment for patients with instrumented hip implants. Bergmann et al. (2001) tested many daily activities including walking, stair ascent and descent. Peak hip contact forces for stair ascent were lower than stair descent but similar to fast walking. The peak AP force was greater for stair ascent than normal walking. The peak vertical forces at the femoral head of normal walking and stair ascent did not differ much. Studies applied instrumented hip implant measured hip loading directly and concisely, but problems still remained if these loading patterns can be applied to healthy populations: 1) some differences in movement patterns due to limitation of hip joint movement, influence of recovering time, and muscle forces loss; 2) the number of subjects selected was so limited, always 2-4 participants in each study.

By testing ground reaction force and free moment and using inverse dynamics and musculoskeletal models, calculation of moments and forces acting at lower limb joints is possible. Edwards et al. (2008) measured ground reaction forces and free moments in running and calculated reaction forces, and moments at lower limb joints as well as internal bone forces and moments on 11 points of the femur. Femoral neck had greater AP force and the lowest axial force, but its peak axial load always occurred during impact phase, which was associated with high loading rate. The loads of femoral neck during running were shown, but the effect of other activities, e.g. stair ascent and

descent, on femoral stress distribution was not taken into consideration. Kirkwood et al. (1999) measured hip moments for older population. Peak internal moments were measured, which showed that most internal hip moments for stair ascent, descent and walking didn't differ much. This conclusion was contrary to the results from Bergmann et al. (1995), which showed that bending moment at hip were greater for stair ascent and descent than walking at 3 km/h, but comparable to walking at 5 km/h.

Based on above comparisons, the outcomes could be varying if researchers represented loads via moments or via contact forces. Recalling the analysis of materials in engineering, stress analysis can be a better method to analyze the loads at the femoral neck, since both of the total loads and bone structure are taken into consideration. The purpose of this study was to compare stresses on the femoral neck during stair ascent and descent. The hypothesis is that loading at the femoral neck will be greater for stair ascent than descent. Patterns of femoral neck stress were also compared to the loading patterns from sagittal and frontal plane internal hip moments to see if there is agreement between the femoral neck stress and hip moment patterns.

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CHAPTER 2. LITERATURE REVIEW

Fractures are among the most serious injuries for people, and the causes of fracture can be various. Stress fracture is caused by repetitive loading at the bone (Burr et al., 1985), while other fractures could be due to various reasons: activities with extremely high load (Turner et al. 2005), the loss of BMD (McCreadie et al. 2000), and a variety of illnesses.

For people, especially for older ones, hip fracture is one of the most common types of fracture and among the injuries of most seriousness. For females, the incidence of fracture of the femoral neck doubles every 5 years after the age of 60, and the average risk at age 80 is about 7 percent (Jensen et al. 1980). For all stress fracture cases, almost half of stress fractures occur on the femoral neck (McBryde et al. 1985; Niva et al. 2005), such fractures at the femoral neck are also among the most serious of fracture injuries (Lord et al. 1986). Fractures at the femoral neck play an important role in morbidity and mortality among people, especially among older individuals (Graves et al. 1992). Great numbers of hip fractures, including femoral neck fractures, lead to more and more disability or death every year (Cummings et al. 1992; Kenzora et al. 1984; White et al. 1987). Moreover, healing of hip fractures costs a greater amount of health funding in the United States, with estimated amounts per year about $7.1 billion (Praemer et al. 1992). These statistics show the importance of exploring the mechanism of femoral neck fractures and how to avoid such fractures in daily life.

As above mentioned, high load, repetitive or non-repetitive, is a crucial risk factor f or such fractures. To explore the mechanism of femoral neck fractures, the load conditions at the femoral neck in different activities (Bergmann et al. 1995) need to be known, including stresses, internal bone forces and moments at the femoral neck, and hip joint reaction forces and moments.

To explore the loading conditions of the femoral neck, many studies focused on cadaver testing. One study carried out by Keyak et al. (2001) used automatically generated, CT scan-based linear finite element models of proximal femora to calculate the force directions that related to the lowest fracture loads and related their data to hip fracture risk. Four right proximal femora from donors whose age were more than 55 were selected. These femora were scanned by CT and 3 dimensional finite element models were generated based on CT scans. These models were evaluated by two types of loading: fall condition whose force was applied to the femoral head at an angle γ to the sagittal plane and an angle δ to the frontal plane, and atraumatic loading condition whose force was applied to the femoral head at an angle α to the sagittal plane and an angle β to the frontal plane. The results showed that under the fall condition the minimum average fracture load of 1121 N occurred at γ= 70 degrees and δ=55, and the maximum average fracture load of 1797 N occurred at γ=80 and δ=85. Under atraumatic loading condition, the lowest fracture loads occurred when β=30 or when α= 10 and β=0, and the highest loads occurred at α=20 and 25, β=7 and 10. But the researchers didn’t point out if these conditions could be indications of the real landing or standing conditions during daily activities, so whether the force direction applied to the femoral head during daily

activities is similar to the direction which leads to lowest fracture loads of hip joint is still in doubt. Moreover, this study didn't take muscles across femoral neck into consideration. Some hip abductor muscles could be protective to femoral neck in daily activities, but cadaver testing could not reflect the effect of this protection.

In another study, Swiontkowski et al. (1987) selected 19 matched pairs and 1 unmatched pair of proximal femora from people ranging in age from 48 to 90 years old. These proximal femora were loaded by torsional testing, which applied an 890 N compression load along the femoral neck axis and the shaft was rotated around the femoral neck axis at the rate of 270 degrees/min. Then these femora were loaded by a cyclic bending test which cycled at 1 Hz, with the vertical compression load at femoral head ranging from -222 to -666 N, -445 to -1355 N, and -667 to -2000 N (negative was downward direction). Researchers measured the average bending and torsion parameters from these specimens: bending stiffness for femoral neck were 2.005 MN/m (SD=0.673, loading ranges from -222 N to -667N), 2.174 MN/m (SD=0.609, loading ranges from -445 N to -1335 N), 2.179 MN/m (SD=0.543, loading ranges from –667 N to -2000 N). Torsional stiffness was 378.4 Nm/rad (SD=111.2). However, this study has the same problem with Keyak's study (2001), since analysis for applied forces during real daily activities was still not applied to their cadaver limb study.

Some other studies worked on measuring contact forces and moments at hip joint during different daily activities directly, by using instrumented hip implants with telemetric data transmission. Bergmann et al. (2001) selected 4 older subjects (age range: 51-76) with instrumented hip implants and asked them to do daily activities including

walking with 3 different speeds (slow: 3.5 km/h; normal: 3.9 km/h; fast: 5.3 km/h), stair ascent and descent. They gathered the contact forces and moments at hip joint. Resultant peak hip contact forces for walking and stair ascent/descent were reported as follows: slow walking was 242% BW, normal walking was 238% BW, fast walking was 250% BW, stair ascent was 251% BW, and descent was 260% BW. The peak contact forces for slow walking and normal walking did not differ much. Stair ascent was lower than descent but similar to fast walking. The peak AP force at femoral head is highlighted in this study since it causes higher implant torque. It was greater for stair ascent (-60.6% BW) than normal walking (-32.8% BW). The main force that had effects on the femoral neck was vertical force acting at femoral head, which produced bending at the femoral neck. The peak vertical forces at femoral head were shown for normal walking (-229.2% BW) and stair ascent (-236.3% BW), but the difference was small. The moments acting on the hip joint were also reported. The torsional moments for walking were 1.64% BW m , 1.52% BW m, and 1.54% BW m, for stair ascent and descent were 2.24% BW m and 1.74% BW m. Since the purpose of this study was to improve hip implants, researchers focused more on torsional moment in transverse plane and AP force at the femoral implant instead of ML moment and vertical force at hip or femoral neck which are considered as main factors of stress fracture.

In another study, Heller et al. (2001) examined musculoskeletal loading at hip during walking, stair ascent and descent. Four total hip arthroplasty patients (age range: 51-76) were included in this study, and all of them had instrumented femoral prostheses which were used in measuring the in vivo hip contact force by Bergmann et al. (1988,

1993, 1995). The researchers used inverse dynamics and musculoskeletal model to calculate hip loading and compared calculated data with those measured directly by instrumented femoral prostheses. Such each-trial comparisons showed that there was a good agreement in pattern and magnitudes of calculated and directly measured hip contact forces for walking in all 4 subjects. The mean of relative deviation (the difference between measured and calculated force divided by measured force) was 12%. For stair ascent and descent, the calculated forces also agreed well with the measured data, the mean of relative deviation was about 14%. For walking and stair ascent/descent, a small overestimation of the hip contact forces occurred in calculation. Although the researchers argued that walking and stair ascent/descent were 2 daily activities with both large number of load cycles and large hip contact forces, they did not report the difference between calculated or measured loading for walking and for stair ascent/descent. As a result, we could not tell which activity in this study generated greater load at hip joint and had more effect on load at the femoral neck.

Bergmann et al. had two other studies that involved instrumented hip implants with telemetric data transmission and hip loading. The first study (1995) hypothesized that stair ascent was an activity which caused greater hip joint forces and moments. Measurements were taken in 2 subjects (age: 69, 82), forces and moments at hip joint were obtained during stair ascent/descent and compared to those during level walking obtained by other studies. For stair ascent at normal speed, the hip joint contact force (subject 1: 350% BW; subject 2: 552% BW) was much greater than during walking with the speed of 3 km/h (subject 1: 315% BW; subject 2: 409% BW). Stair descent also

increased hip joint contact forces (subject 1: 392% BW; subject 2: 509% BW). Hip contact force for walking with the speed of 5 km/h (subject 1: 381% BW) was comparable to stair descent (subject 1: 392% BW), but greater than during stair ascent (subject 1: 350% BW). Both the angle F between the resultant hip contact force and vertical axis and angle T between resultant hip contact force and ML axis were reported. Stair ascent obtained the greatest angle F (26 deg.) and the greatest angle T (29 deg.), which resulted in smaller portion of vertical force and ML force from resultant force. Angle F and T in average was smaller for stair descent (subject 1: 21 deg.) than that for walking at 3 km/h (subject 1: 24 deg.). Angle T in average were similar for stair descent (15.5 deg.) and walking at 3 km/h (14 deg.). Stair ascent generated a torsional moment (4.0% BW m) that was about twice as high as during walking at 3 km/h (2.0% BW m), but walking at 5 km/h (3.4% BW m) and slow jogging (4.3% BW m) caused torsional moments at the similar level with stair ascent and greater than descent (2.6% BW m). The bending moments were similar for stair ascent (7.0% BW m) and descent (8.3% BW m), while a little greater than walking at 3 km/h (6.4% BW m). Bergmann et al. (1995) also examined the influence of shoes and intensity of heel strike on forces and moments at the hip joint during walking at 3 km/h and jogging at 6 km/h. Walking without shoes caused resultant hip contact force of 289% BW, with most shoes the values of resultant force were slightly greater, but the changes (-2 to +6%) were small. These values were much smaller than those for stair ascent/descent in the last study. The bending moments for barefoot was 4.8% BW m, and the increase of bending moment with shoes was relatively uniform, ranging from +4 to +10%. Magnitudes of these values were about

half of bending moment for stair ascent/descent in the last study. Jogging at 6 km/h without shoes generated 472% BW of resultant hip joint force, and all shoes generated slightly greater resultant hip joint forces (+3 to +6%), which were much higher than stair data from subject 1 in the last study. The bending moment for jogging was about 6.8% BW m with barefoot, and wearing shoes ranged from -4 to +6%, which was similar with stair descent but a little higher than stair ascent. Bergmann et al. also introduced loading rate, an important concern in studies of bone loading. Walking with different shoes caused loading rate from 2235% BW/s to 3076% BW/s, and jogging with different shoes generated loading rate from 5597% BW/s to 8452% BW/s. Since there was no loading rate data for stair ascent/descent, the comparison between stair ascent/descent and walking or jogging could not be preformed.

Although studies with instrumented hip implants measured hip loading directly during many activities, problems still remained if the loading patterns of different activities could be used for healthy populations. Firstly, there might be some differences in movement patterns between injured subjects and healthy ones. Limitation of hip joint movement, influence of recovering time, and muscle force loss would change their patterns of various movements, and their data may not be appropriate to apply to healthy population. In one of Bergmann's studies (1995), variance between the outcomes of his 2 subjects was quite great, e.g. the hip joint contact force for stair ascent was 350% BW in subject 1, 552% BW in subject 2; and for stair descent was 392% BW for subject 1, 509% BW for subject 2. Moreover, the number of subjects selected in each study was so small, and whether those small groups of subjects could represent the characteristics of

the hip implant population is still in doubt. For example, Bergmann et al. (1995, 2001) selected 2-4 subjects in his studies and Heller et al. (2001) only had 4. It was hard to get more subjects with instrumented hip implants in their studies, and this situation limited ability of generalization for such studies.

Indirect methods were also used in some studies focused on determining the load at femoral neck or hip joint. The researchers measured ground reaction forces and free moments for a variety of activities and applied inverse dynamics and musculoskeletal model to figure out the loading at the joint, which made calculation of moments and forces acting at lower limbs possible. One study carried out by Edwards et al. (2008) measured ground reaction forces and free moments during running in young male runners. Joint reaction forces, moments at ankle, knee and hip as well as internal forces and moments were calculated to study internal load on multiple points of the femur and the relation to stress fracture condition during running. In this study of 10 male runners, anthropometric data, motion data and ground reaction forces & free moments were collected. A scaled SIMM musculoskeletal model and fmincon function in Matlab were used to obtain forces and moments at the femur generated by lower limb muscles. Internal forces and moments were calculated along a centroid path at 11 equidistant points within the femur. Joint contact forces at 3 lower limb joints, internal femoral forces and moments at those 11 points were also reported. Peak AP joint contact force at the hip joint was -1.60±0.45 BW, which was smaller than the knee (-1.83±0.08 BW). Peak axial force at the hip was -11.89±2.19 BW, which was also much smaller than the knee (-15.09±0.59 BW). The peak ML force at the hip was 6.25±0.83 BW, and the

absolute value was greater than the knee (-1.19±0.07 BW). The femoral neck peak AP force was 3.57 BW, peak axial force was -6.79 BW, and peak ML force was -3.75 BW. Peak AP moment for the femoral neck was -0.31 BW m, peak torsional moment was 0.05 BW m, and peak ML moment was -0.20 BW m. Femoral neck had greater AP force than other points except the most distal point, which also had greatest absolute ML force. Femoral neck had the lowest axial force, but the peak load occurred during impact phase of running, which was associated with high loading rate that might lead to more micro-damages to the bone tissue. The moments at femoral neck were not much different with the average of all 11 points. In this study, researchers found out the loading conditions of femoral neck during running for young runners, but other activities, such as stair ascent/descent, were not taken into consideration.

Kirkwood et al. (1999) measured hip moments during level walking, stair ascent/descent and other exercises in an older population (30 subjects 55 yrs or older) with no identified musculoskeletal or neurological impairment. Their purpose was to figure out which exercises resulted in the greatest moment at hip joint. They collected motion, ground reaction force and anthropometric data for more than 10 types of activities, including level walking, stair ascent and descent. Hip moments were calculated by the use of inverse dynamics with a link segment model. They reported peak internal moments in 3 different planes for 14 activities. In frontal plane, the maximum mean peak internal hip abductor moment was generated during stair descent and the value reached 0.96 Nm/kg, while the mean value during level walking was 0.91 Nm/kg. Smaller hip abductor moment (0.77 Nm/kg) was generated during stair ascent

than level walking (0.82 Nm/kg). However, the moment differences (level walking vs. stair ascent and level walking vs. stair descent) were not significant. The internal hip adductor moment collected during level walking was much greater than most of the other exercise. Stair ascent and descent generated internal hip adductor moment of -0.07 Nm/kg and -0.04 Nm/kg, which were significantly smaller than -0.18 Nm/kg in level walking.

In sagittal plane, no exercises reached higher peak internal hip flexor moment than those in level walking. Peak internal hip flexor moment during stair ascent reached 0.20 Nm/kg, much smaller than level walking (0.78 Nm/kg). Mean hip flexor moment was 0.28 Nm/kg during stair descent and was smaller than level walking (0.68 Nm/kg). Differences between level walking and stair activity were significant. The maximum hip extensor moments obtained for stair ascent (-1.0 Nm/kg) and level walking (-0.94 Nm/kg) were not significantly different. Stair descent (-0.50 Nm/kg) generated only half the amount of internal extensor moment during level walking (-1.03 Nm/kg).

In the transverse plane, the hip external rotation moments generated by stair ascent were not much different from those obtained during level walking (0.10 Nm/kg vs. 0.08 Nm/kg). Similarly, hip external rotation moments from stair descent were comparable to those from level walking (0.09 Nm/kg vs. 0.09 Nm/kg). Hip internal rotation moments obtained for stair ascent (-0.21 Nm/kg) were about as two times as high as those obtained during level walking (-0.11 Nm/kg). Stair descent (-0.14 Nm/kg) generated similar value of hip internal rotation moment to level walking (-0.12 Nm/kg).

Rate of moment changing during those 14 activities were also shown. In frontal plane, stair descent had a rate of moment changing comparable to level walking (0.111 Nm/kg/s vs. 0.067 Nm/kg/s). Stair ascent required much smaller moment changing rate than level walking (-0.060 Nm/kg/s vs. 0.096 Nm/kg/s). In sagittal plane, both stair ascent (-0.090 Nm/kg/s) and descent (-0.081 Nm/kg/s) had much lower moment changing rates than level walking (-0.166 Nm/kg/s compared to stair ascent in one group, -0.179 Nm/kg/s compared to stair descent in another group). In transverse plane, the rates of moment changing for stair ascent (-0.020 Nm/kg/s) and descent (-0.009 Nm/kg/s) did not differ too much from those obtained during level walking (-0.017 Nm/kg/s).

This study showed that most internal hip moments during stair ascent and descent were not much different from level walking. The moments which had more influence on femoral neck fracture were those in frontal plane. The results show that there was no significant difference in hip abductor moment between level walking and stair ascent or stair descent, while adductor hip moments during both stair ascent and descent were significantly lower than those during level walking, which meant that stair ascent/descent might not generate higher load on the hip. This conclusion might be contrary to the part of the results from Bergmann et al. (1995), which showed that bending moment at hip in frontal plane were larger during stair ascent/descent than level walking at 3 km/h, but comparable to during level walking at 5km/h. Since Kirkwood et al. (1999) did not show the magnitude of walking speed, a comparison in detail could not be presented.

Some of the above studies used indirect measurements in calculating the loading at femoral neck or hip joint. Ground reaction forces during activities such as walking and stair ascent/descent were measured by force platform. Retro-reflective markers were put on anatomical landmarks of the trunk and one side lower limb, and motion-capture data were collected by 3-D capture system. Combining force data and motion data with anthropometric data, forces and moments of the lower limb joints could be calculated by inverse dynamics. Theoretically, this calculation was accurate for more distal joint, e.g. ankle joint, but could result in errors for more proximal joint, e.g. hip joint, due to more steps of calculation (Heller et al. 2001). However, Heller et al. (2001) compared the hip loading results calculated by inverse dynamics method and musculoskeletal model with those measured directly by instrumented femoral prostheses during walking and stair ascent/descent. Even though there was a small overestimation of the hip contact forces for both walking and stair ascent/descent (no more than 13%), a good agreement in patterns and magnitudes of calculated and directly measured hip contact forces was obtained for walking and stair ascent/descent.

The inverse dynamics calculated load at joints, but it was not enough to reveal the loading distribution for the whole bone or on one or more particular parts of bone tissue. To get such information, external and internal loads should be considered together. For example, if femur needs to be examined, we should consider both the loads from joint and forces generated by muscles attached to the femur. One musculoskeletal model researchers use most frequently is the OpenSIMM model. This musculoskeletal model can obtain dynamic maximal muscle forces, moment arms, muscle attachment

sites, and muscle orientations for all 43 muscles for the lower limb. All maximum dynamic muscle forces were adjusted for muscle length and velocity for individual participants.

When we calculated maximum muscle forces and moments for each lower limb muscle, we guessed a set of muscle forces for each muscle, and the sum of muscle moments should equal joint moments calculated by inverse dynamics. Many possible sets of muscle forces are created by this equation, and we choose the set of muscle forces that minimizes the cost function. We use the fmincon function in Matlab to optimize the muscle forces (Glitsch et al. 1997). Joint contact forces are calculated when the net force at the joint plus all muscle forces related to the same bone are summed.

In some other studies, finite element model was used in calculating loading at the bone tissue. The bone specimens were CT scanned and a three dimensional linear finite element model was generated from CT data of those specimens as 3-mm (this value differs if different CT scanners are used) linear cube-shaped elements (Keyak et al. 2001). In these models, elastic modulus and compressive strength could be computed in each element of the bone by using correlations between calibrated CT scan data and ash density, and between ash density and mechanical properties of trabecular and cortical bone. One major usage of these models was testing loading types at the bone which represented some conditions like standing, walking or landing. Most studies applied these models aimed at finding the loading conditions and failure criteria of various parts of bone, testing different failure theories for the bone (Keyak et al. 2001, Sabick et al. 1997). Some studies showed that the finite element method has a good accuracy in

predicting bone strength and bone fracture load. Cody et al. (1999) compared the femoral strength prediction made by finite element models and such predictions made by quantitative computed tomography (QCT) and dual energy X-ray absorptiometry (DXA). The finite element method explained greater than 20% more variance in the predicted fracture load than did the DXA method, while QCT was predicted better than DXA but not as good as finite element method. Keyak at el. (2001) also found significant relationships between measured fracture load and finite element predicted fracture load (r=0.87 for stance, r=0.95 for fall, r=0.97 for stance and fall pooled). However, there is a problem of using this method: most subjects in the finite element model studies were older individuals. The cadaver material in the study of Keyak et al. (2001) were from donors aged from 52 to 92 years old, and Cody et al. (1999) selected 51 donors all of whom were older than 42 years old. The data from this finite element model may not be applicable to bone studies involved young healthy people, since bone geometry, bone density and material properties may not be the same.

Another method is based on the software, VA-BATTS (Kourtis et al. 2008). It helps to analyze the elastic behavior of long bones under axial, bending, torsional and transverse loading conditions. This analysis uses subject-specific geometric and material data from CT images (usually the cross sectional image of the long axis of the bone) and imports and these images into a 2-D automated finite element analysis routine generated by a QCT image, to determine the stresses at the bone. This method is more accurate than 2-D models of idealized bone cross sections, and less complicated than 3-D models which are suitable for geometrically complex structures. Kourtis et al. (2008) reported

that there was an excellent agreement between this model and 3-D finite element model, with the differences averaged over all of elements ranging from 1.2 to 1.4% and differences in peak values ranging from 0.3 to 1.7%.

One purpose of this study was to compare stresses on the femoral neck during stair ascent and descent. The hypothesis for this purpose is that: loading at the femoral neck will be greater for stair ascent than descent. Moreover, based on previous studies both sagittal and frontal hip moments are greater in stair ascent than descent (Novak et al. 2011; Riener et al. 2002), but it is unknown if stresses at the femoral neck have a similar pattern. In this study, patterns of femoral neck stress were also compared to the loading patterns from sagittal and frontal plane internal hip moments to see if there exists any agreement or disagreement between the femoral neck stress and hip moment patterns.

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CHAPTER 3. MEASURING FEMORAL NECK LOADS IN OLDER ADULTS DURING STAIR ASCENT AND DESCENT

Introduction

Fractures are among the most serious injuries for people, but the causes can vary. Repetitive loads, high magnitudes (Turner et al. 2005), the loss of bone mineral density (BMD) (McCreadie et al. 2000), and a variety of illnesses may be the main cause of a fracture.

Fractures at the femoral neck play an important role in morbidity and mortality among people, especially older individuals (Graves et al. 1992). For females, the incidence of fracture of the femoral neck doubles every 5 years after the age of 60, and the average risk at age 80 is about 7 percent (Jensen et al. 1980). To explore the mechanisms of femoral neck fractures, the loads at the femoral neck in different activities, including stair ascent/descent and running, need to be known (Bergmann et al. 1995). The loads could be presented in the form of stresses, forces or moments at the hip joint or femur.

To investigate the loading at the femoral neck for different activities, we need to determine reaction forces and moments as well as muscle forces at the hip joint. Keyak et al. (2001) studied the cadaver proximal femora using CT scan-based linear finite element models. Two types of loading were used: a fall and atraumatic loading conditions. The results showed the values and directions of minimum and maximum

fracture load for both conditions. However, the results did not show if the direction of the applied force during daily activities were similar to their settings. This problem was also evident in another study by Swiontkowski et al. (1987), in which the average bending and torsion parameters were reported through material testing of cadaver proximal femora.

Several studies directly measured hip contact force and moment for patients with instrumented hip implants. Bergmann et al. (2001) tested many daily activities including walking and stair ascent/descent. Peak hip contact forces for stair ascent were lower than stair descent but similar to fast walking. The peak anterior-posterior (AP) force was greater for stair ascent than normal walking. The peak vertical forces at the femoral head for normal walking and stair ascent were not significantly different. For studies measuring hip loads directly with instrumental hip implants, problems may arise when applying these loading patterns to healthy populations: 1) differences in movement patterns due to limitations in hip joint movement, influence of recovery time, and muscle forces loss; 2) the number of subjects selected, 2-4 participants in each study.

Calculation of moments and forces acting at lower limbs joints is possible by measuring ground reaction forces and free moments and using inverse dynamics and musculoskeletal models. Edwards et al. (2008) calculated reaction forces and moments at lower limb joints as well as internal bone forces and moments on 11 points of femur during running. The femoral neck had the greatest AP force and the lowest axial force, while its peak axial load always occurred during the impact phase, which was associated with a high loading rate. The loads of femoral neck during running were established, but

the effect of other activities, e.g. stair ascent and descent, on femoral stress distribution was not investigated. Kirkwood et al. (1999) measured hip moments for an older population. Peak internal hip moments during stair ascent and descent were not significantly different. This conclusion was contrary to the results from Bergmann et al. (1995), which showed that the hip moment was greater for stair ascent/descent than walking at 3 km/h, but comparable to walking at 5 km/h.

Based on the above comparisons, the outcomes could vary if researchers represented loads via moments or via contact forces. Using mechanics of materials, stress analysis can be a better method to analyze the loads at the femoral neck, since both the total loads and bone structure are taken into consideration. The purpose of this study was to compare stresses on the femoral neck during stair ascent and descent. The hypothesis was that loading at the femoral neck would be higher for stair ascent than descent because of the increased muscle forces required during the ascent activity. Patterns of femoral neck stress were also compared to the loading patterns from sagittal and frontal plane internal hip moments to see if there is agreement between these two types of analyses.

Methods

Subjects

Ten subjects (5 males and 5 female, aged from 50 to 70) volunteered to participate in this study. All of the participants were free from lower limb injuries during data collection. Before participation, they signed a written informed consent document that had been approved by the Iowa State University Human Subjects Review Board.

Data collection

A series of anthropometrics were measured for each subject, including total mass, thigh length, midthigh circumference, calf length, calf circumference, foot length, malleolus height and malleolus width. Eighteen reflective markers were placed on anatomical landmarks of the trunk, pelvis and right lower limb: head of the fifth metatarsal, dorsi-foot, heel, medial and lateral malleoli, lateral calf, posterior calf, medial and lateral femoral epicondyle, anterior and lateral thigh, both greater trochanters, both anterior superior iliac spines (ASISs), both posterior superior iliac spines (PSISs) and sacrum. All anthropometric measurements and reflective marker placements were performed by the same researcher. A static trial was collected to estimate joint center locations and reflective markers on the medial leg were removed prior to further testing.

All subjects performed 5 successful trials of stair ascent and same number of trials of descent (a 3-level staircase, the height of each stair is 19 cm). The left foot started each trial and the right foot contacted the force platform on the second step. Two AMTI force platforms (1200 Hz, AMTI, Watertown, MA) were placed on 2 lower stairs

to gather ground reaction force data. Motion data were collected by a Vicon system (120 Hz, Vicon MX, Vicon, Centennial, CO, USA).

Data analysis

Custom software was used for further processing and analysis of data (Matlab). The ground reaction forces were decimated to 120Hz and smoothed with a low-pass cutoff frequency of 6Hz. The motion data were smoothed with same method with a cutoff frequency of 6Hz. The stance phase cycle for stair ascent/descent began with right foot first contact on the force platform, and finished with toe-off of same foot on the same force platform. All the cycles were normalized into a percentage of this stance phase.

Segment masses, center of mass locations, and moments of inertia were obtained according to Vaughan et al. (1992) by using the anthropometric data input into the Matlab program. Using inverse dynamics and rigid body assumptions, joint moments and reaction forces were calculated for the ankle, knee and hip joint (equations for ankle below):

$$R_p = ma_{cm} - R_d - mg$$

$$M_p = I\alpha - M_d - (d - CM) \times R_d - (p - CM) \times R_p$$

Where, R_p is the proximal reaction force
m is the mass of the segment
a_{cm} is the acceleration of the center of mass of the segment
R_d is the distal reaction force (GRF for the foot)
g is [0, -9.81, 0]
M_p is the proximal moments
I is the moments of inertia for the segment
α is the angular acceleration of the segment

M_d is the distal moments ([0, free moment, 0] for the foot)
d is the coordinates of the distal end of the segment (COP for the foot)
CM is the coordinates of the center of mass of the segment
p is the coordinates of the proximal end of the segment

Moments and reaction forces were calculated in the global coordinate system and then transformed into the coordinate system of the proximal segment at each joint. Cardan angles for each joint were calculated using a flex-ext/abd-add/int-ext rotation order of rotations. These angles were input into a musculoskeletal model using the joint and muscle definitions of Delp (1990). The musculoskeletal model was implemented in Matlab. Dynamic length and velocity adjusted maximal muscle forces, muscle moment arms and orientations for 43 lower limb muscles were obtained in this program. A static optimization was used to select a set of muscle forces that minimized the sum of the squared muscle stresses and balanced the sagittal plane hip, knee and ankle moments and the frontal plane hip and ankle moments for each frame of data. Joint reaction forces for the hip and knee were summed with the muscle forces to obtain joint contact forces. Joint contact forces were referenced to the local coordinate system of the distal segment so that hip contact forces represented the external forces acting on the head of the femur.

Forces at the centroid of the femoral neck cross-section were calculated by summing the reaction and muscle forces and then transforming into a femoral neck coordinate system. Moments at this site were calculated using the hip joint moment and the moments generated by the hip reaction force and the muscles that cross the hip (Duda et al. 1997).

An ellipse bone model was used to estimate stresses on the surface of 4 sites on the femoral neck: superior, inferior, anterior and posterior. The superior-inferior and anterior-posterior diameters of the model were 3.6 and 2.5 cm according to Alunni-Perret et al. (2003) and Gnudi et al. (2002). The superior-inferior (SI) and anterior-posterior (AP) thicknesses of the model were 0.6 and 0.3 cm according to Bell et al. (1999), based on the cortical bone thickness estimates of the femoral neck. This model had a cross-sectional area of 2.576×10^{-4} m^2 and moments of inertia about the AP axis of 3.827×10^{-8} m^4, and about SI axis of 1.402×10^{-8} m^4. Stresses were estimated using the following formulas:

$$\sigma_{\mathrm{superior}} = \sigma\left(-\mathrm{M}_{\mathrm{ap}}\right) + \sigma(\mathrm{F}_{\mathrm{axial}})$$

$$\sigma_{\mathrm{inferior}} = \sigma\left(\mathrm{M}_{\mathrm{ap}}\right) + \sigma(\mathrm{F}_{\mathrm{axial}})$$

$$\sigma_{\mathrm{anterior}} = \sigma(\mathrm{M}_{\mathrm{ml}}) + \sigma(\mathrm{F}_{\mathrm{axial}})$$

$$\sigma_{\mathrm{posterior}} = \sigma(-\mathrm{M}_{\mathrm{ml}}) + \sigma(\mathrm{F}_{\mathrm{axial}})$$

Where $\sigma_{\mathrm{superior}}$ is the stress on the superior aspect of the femoral neck, $\sigma_{\mathrm{inferior}}$ is the stress on the inferior aspect of the femoral neck, $\sigma_{\mathrm{anterior}}$ is the stress on the anterior aspect of the femoral neck, $\sigma_{\mathrm{posterior}}$ is the stress on the posterior aspect of the femoral neck, $\sigma(\mathrm{M}_{\mathrm{ml}})$ is the stress generated by ML moment, $\sigma\left(\mathrm{M}_{\mathrm{ap}}\right)$ is the stress generated by AP moment and $\sigma(\mathrm{F}_{\mathrm{axial}})$ is the stress caused by the axial force.

Statistical analysis

The main dependent variables in this study were stresses at the 4 sites on the femoral neck. Differences in stresses for stair ascent and descent were examined by

dependent t-test. All statistical tests were considered significant at $p<.05$. Statistical analyses were performed in IBM SPSS Statistics 19.

Results

The sample for ten older adults was composed of five males and five females. The average age, body mass, and height are shown in the Table 1.

Table 1. Means (SD) of basic information for all subjects.

	Age(yr)	Body mass(kg)	Body height(m)
	59	68.90	1.68
	(6.1)	(10.30)	(0.05)

Hip joint moments are shown in Figure 1, normalized by body weight. Greater absolute moment values were shown in stair ascent (sagittal: -0.077±0.022; frontal: -0.094±0.011) compared to descent (sagittal: -0.036±0.018; frontal: -0.091±0.015). However, the differences for peak hip moments in frontal plane showed no significance ($p = 0.705$). Peak hip extension moments were significantly increased in stair ascent ($p = .001$).

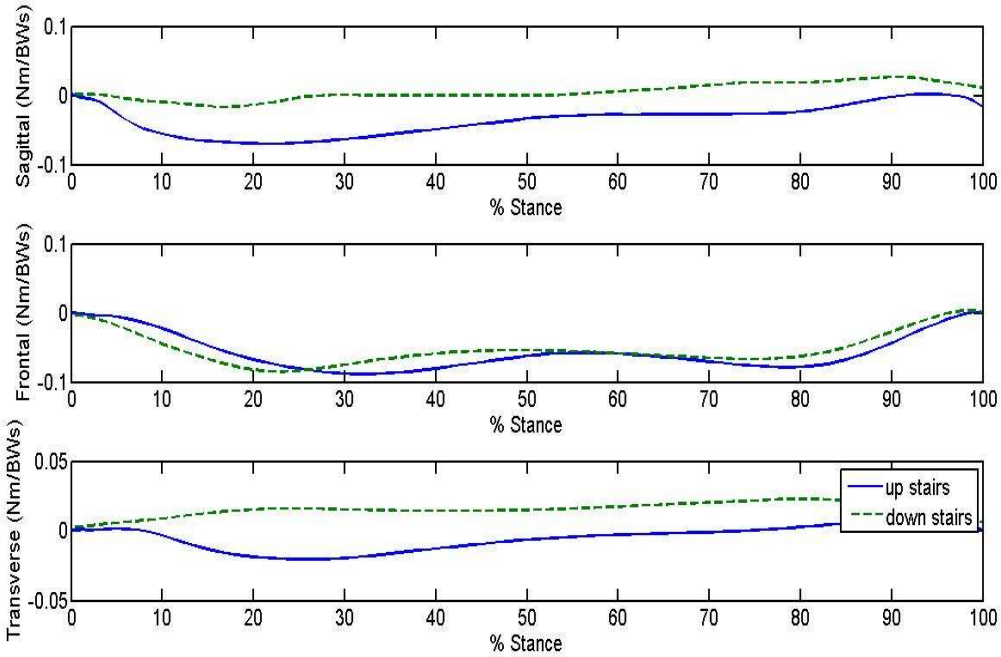

Figure 1. Ensemble average of hip moments in 3 planes. Positive values are flexion, adduction and internal rotation.

Hip contact forces (reaction contact forces at thigh segment) are shown in Figure 2, normalized by body weight and based on thigh coordinate system. Peak medial contact forces ($p = .019$) and peak anterior contact forces ($p = .021$) were significantly increased during stair ascent. Peak vertical contact forces increased during stair descent ($p = .018$). However, when comparing two distinct peaks individually, there was no significant difference in vertical contact forces for the 1st ($p = .263$) and the 2nd peak ($p = .110$).

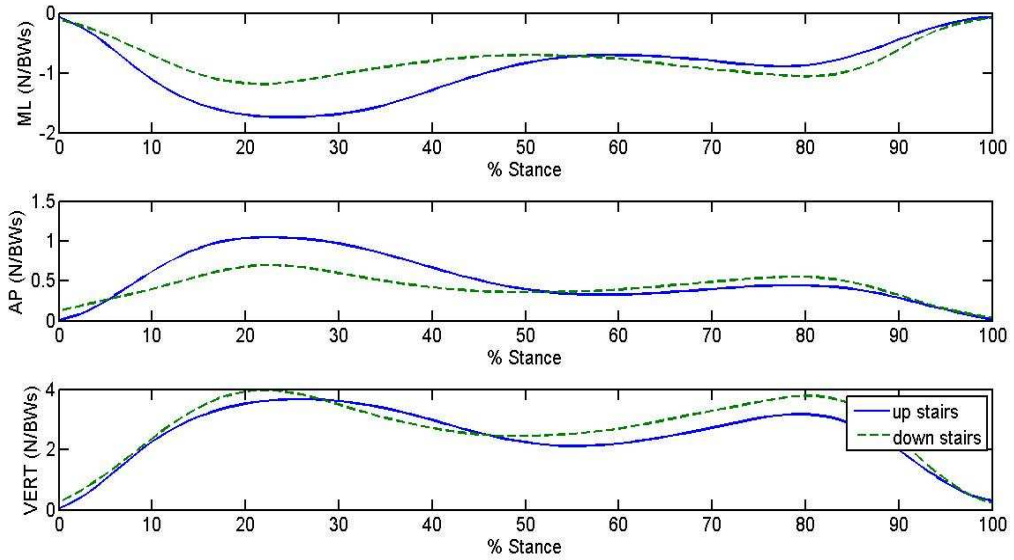

Figure 2. Ensemble average of contact forces at 3 planes of the hip joint. Positive values indicate lateral, anterior, and upward directions.

Estimated muscle forces generated by hip adductor-abductor and flexor-extensor muscles are shown in Figure 3 and normalized by the body weight. For hip extensor and flexor muscles, only gluteus maximus had a significant increase in estimated force during stair ascent ($p < .001$). Long head biceps femoris ($p = .052$), semimembranosus ($p = .424$), and tensor fasciae latae ($p = .168$) had no significant difference between stair ascent and descent. For hip adductor and abductor muscles, gluteus medius ($p = .523$) had no significant difference in estimated force between stair ascent and descent, but adductor magnus had significantly increased estimated force for the 1st half stance during stair ascent ($p = .008$) and for the 2nd half stance during descent ($p = .001$).

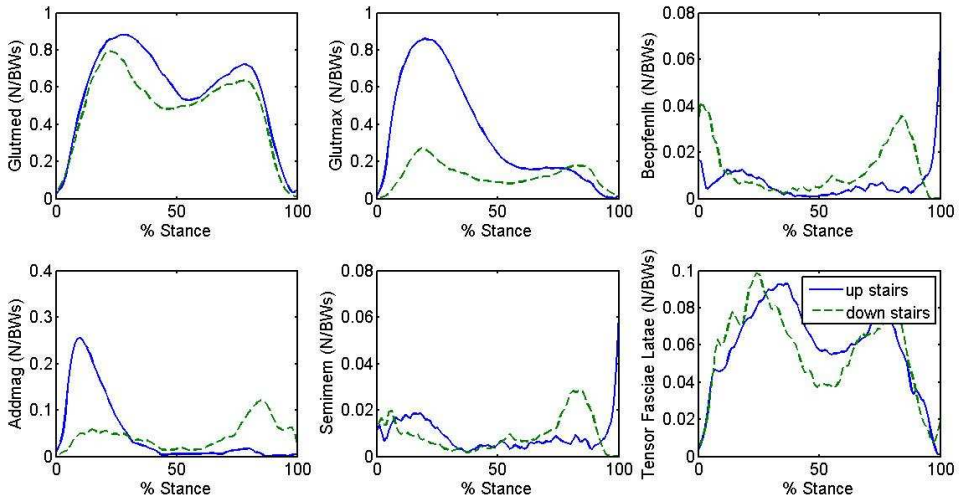

Figure 3. Ensemble average of estimated forces generated by hip adductor-abductor and flexor-extensor muscles, normalized by the body weight. Glutmed = Gluteus medius, Glutmax = Gluteus maximus, Becpfemlh = Long head biceps femoris, Addmag = Adductor magnus, Semimem = Semimembranosus.

Estimated femoral neck stresses and peaks for all 4 sites are shown in Figure 4 and Table 2. Increased peak tension at the superior site and peak compression at the inferior site were found during stair descent. Stresses at both the superior and inferior sites showed distinct peaks during the first half and second half of stance.

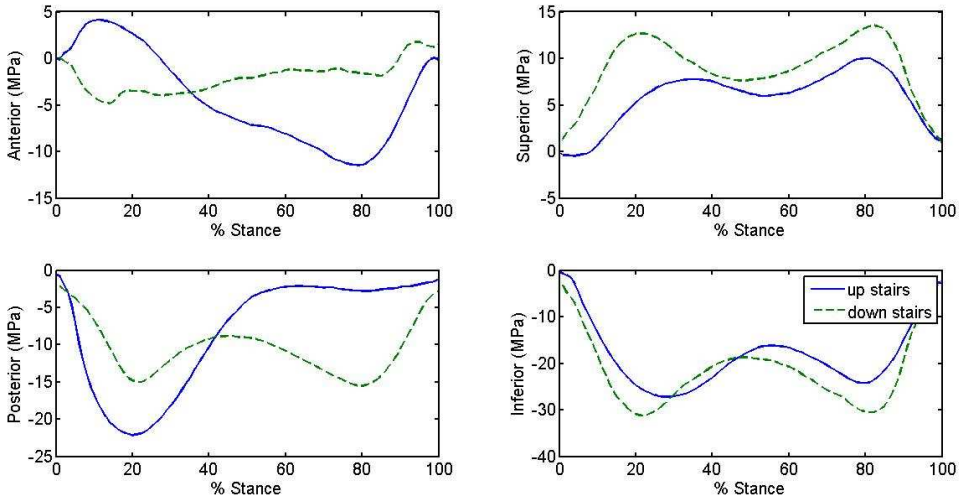

Figure 4. Ensemble average of stresses at superior and inferior sites of femoral neck, positive values indicate tension, negative indicate compression.

Table 2. Means (SD) of peak stresses (in MPa) in 4 sites of femoral neck during stair ascent and descent, P1 indicates peak 1, P2 indicates peak 2.

Stress	Stair ascent		Stair descent	
Sites	P1	P2	P1	P2
Superior	8.4	10.8	13.7	14.4
	(3.4)	(3.9)	(4.0)	(5.4)
Inferior	-28.3	-25.5	-33.1	-32.5
	(6.5)	(6.0)	(6.2)	(9.9)
Anterior	5.0	-12.2	-7.2	-5.5
	(2.6)	(2.4)	(2.4)	(3.6)
Posterior	-23.4	-5.4	-16.0	-16.2
	(5.8)	(2.3)	(5.4)	(4.7)

There was significantly increased peak tensile stress at the superior site for stair descent ($p = 0.005$) in P1, but the P2 showed no significant difference between stair ascent and descent ($p = .098$). Peak compressive stress at the inferior site showed no significant differences in either peak (P1: $p = .105$; P2: $p = .071$) between stair ascent and descent.

At the anterior site of the femoral neck, the 1st peak for stair ascent generated tensile peak stress and descent generated compressive peak stress, which resulted in a significant difference ($p < .001$). Stair ascent generated more compressive stress for the 2nd peak than descent ($p < .001$). At the posterior site, there was a significantly increased 1st peak compressive stress for stair ascent ($p = .009$), and a significantly increased 2nd peak compression for stair descent ($p < .001$).

Discussion

In this study, we estimated the stress conditions in the femoral neck for both stair ascent and descent. The main hypothesis in this study was that loading at the femoral neck would be greater for stair ascent than descent. Moreover, we also planned to determine if stresses in the femoral neck and hip moments share a similar pattern.

When we look at the hip moments, there was a significantly increased hip extensor moment for stair ascent. Insignificant differences in peak hip abductor/adductor moments were also found between stair ascent and descent. These results were in agreement with previous studies which looked at lower limb joint moments for stair ascent and descent conditions (Novak et al. 2011; Riener et al. 2002). The increased hip extensor moment during stair ascent could be explained by more active hip extensor muscles, caused by a more flexed hip posture compared to stair descent. This explanation is also supported by estimated muscle forces in our study and EMG data showed by Lyons et al.1983. Both studies showed increased muscle forces and EMG activity for gluteus maximus, which is a major extensor muscle for the hip joint.

The study hypothesis that stresses would be significantly increased during stair ascent was not supported. The 1st peak tensile stress at the superior site showed a significant increase during stair descent. Several explanations could be made for these outcomes. The 1st peak stress during early stair descent could be explained by the relatively extended position of the hip during decent, which places more of the vertical load in an alignment to create greater tension on the superior neck and compression on the inferior neck through bending stresses. During early stair ascent, the flexed position

of the hip places the vertical loads in an alignment to create tension of the anterior neck and compression of the posterior neck through bending stresses. This trend was also shown clearly in Figure 3, which shows significantly greater peak compression on the posterior site of the femoral neck during early stair ascent, and this increase can be caused by a more flexed hip posture plus greater estimated hip extensor muscle force generated by gluteus maximus.

For the posterior sites of the femoral neck, significantly greater compression peaks were found during late stair descent. This increase can be explained by a greater hip extensor muscle force during late stance of stair descent. Increased extensor muscle force provides tension on the anterior neck and compression on the posterior neck, which partially explains why the anterior neck kept a low level compression during the 2nd half stance.

The patterns of stresses at the superior and inferior sites of the femoral neck were opposite to the hip moments, especially with the hip extensor moment. A figure could be helpful to illustrate these contradict patterns (Figure 5). But the stress patterns at the anterior and posterior sites were more complex since the anterior and posterior sites had completely opposite patterns during 1st and 2nd half of stance phase.

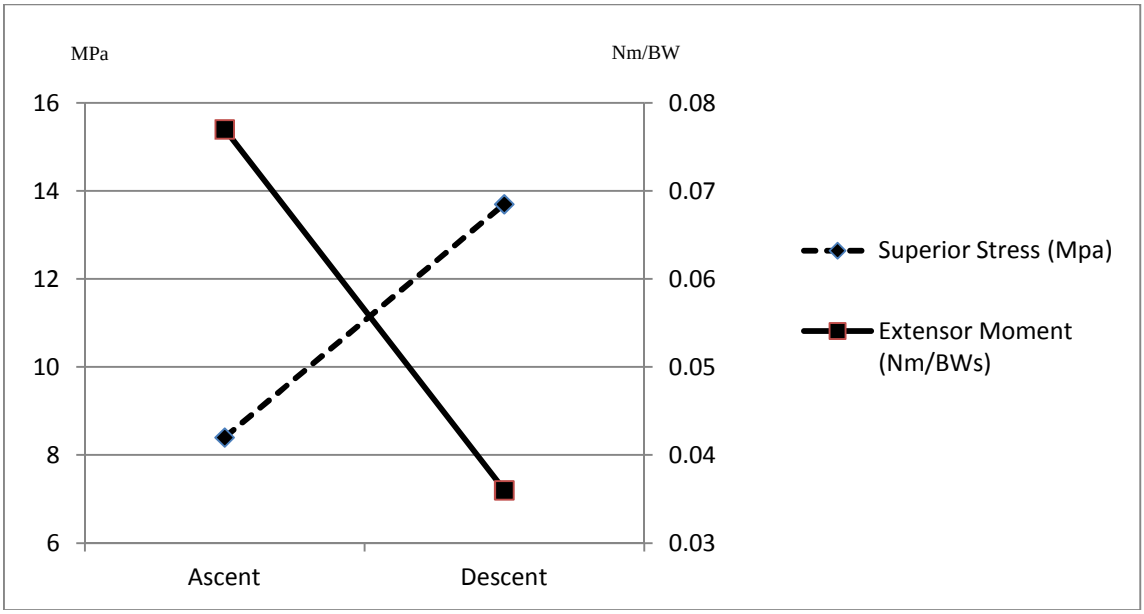

Figure 5. The change of tensile stress on the superior neck and hip extensor moment during stair ascent and descent.

Conclusions concerning the loading of the proximal femur were contradictory depending on if loading was assessed via hip joint moments or from femoral neck stresses. Thus, muscular loading as assessed by the joint moment appears to have an inverse relationship with the skeletal loading as assessed by the bone stress. Such increased femoral neck stresses could be helpful to explain why some older adults have reported more hip pain in stair descent than ascent despite their increased hip moments for stair ascent condition. In such situations, bone tissue might receive relatively greater loading and produce more pain in stair descent, this situation could be due to both more extended lower limb and increased impact peak. Future studies could benefit from a more comprehensive evaluation of the loading environment by estimating bone stresses as well as joint moments during stair ascent and descent. In fact, combining bone

stresses and joint moments could help future studies analyze loading conditions in a more comprehensive way for many other daily or physical activities.

The hip contact forces increased in the AP and ML directions and decreased in the vertical direction during stair ascent. This was in good agreement with previous studies (Bergmann et al. 1988; 1993; 1995). The greater AP contact forces during stair ascent could be explained by increased hip extensor muscle force due to more flexed lower limb position, and greater ML forces could be explained by increased hip abductor muscle force, both of these explanations could be supported by greater estimated hip extensor and abductor forces during stair ascent. However, vertical contact force was greater during stair descent but the difference was not significant.

Estimated hip extensor and abductor muscle forces were also compared to electromyography (EMG) data during stair ascent and descent from previous study (Lyons et al, 1983), but not all muscles showed agreement between estimated forces and (EMG) activities. For stair ascent, semimembranosus, adductor magnus, gluteus maximus, gluteus medius, and tensor fasciae latae showed similar pattern between estimated muscle forces and EMG, while long head biceps femoris showed no agreement in muscle force and EMG patterns. For stair descent condition, only gluteus maximus showed better agreement between estimated force and EMG, but such agreement is weaker than during stair ascent.

Based on the comparisons, large extensor muscles show better agreement than smaller ones. Both force and EMG for gluteus maximus are greater in stair ascent than descent, which show a good agreement with hip extensor moment in our study. Greater

muscle forces do not always indicate greater EMG activities, because EMG is a measure of the stimulation of the muscle force indicates the degree of activation.

However, some limits still remained in our study. One limit in the ellipse bone model we used in the analysis is that: although the shape of this model resembled a real bone structure in general, it's the model with homogeneous material properties while actual bone density is not homogeneous. Another limit for this model is its parameters; we applied averaged femoral neck parameters, including superior-inferior and anterior-posterior diameters, thickness of cortical bone tissues, to the model instead of their individual parameters from computed tomography (CT) or magnetic resonance imaging (MRI) scans. The third limit is the speed on the staircase. In the data collection process, we did not restrict the speed or step frequency for either stair ascent or descent conditions. Differences in speed or step frequency may have influence on the contact forces and hip moments (Bergmann et al. 1995), but how great the influence is still remains to seen. Future work could focus on development of a bone model more consistent with the individual subject or specific age group, CT or MRI scans could be applied to estimate material properties, and the finite element method could be applied to estimate stresses.

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CHAPTER 4. GENERGAL CONCLUSION

The hip moments, hip contact forces and stresses in the femoral neck were estimated for both stair ascent and descent. Looking at loads in different aspect can be beneficial for people to understand the load mechanism during stair ascent and descent.

For the hip moments and contact forces, there was a significantly increased extensor hip moment during stair ascent, and the hip contact forces significantly increased in the AP and ML directions during stair ascent. For the stresses at the superior and inferior sites of the femoral neck, the 1st peak tensile stress at the superior site showed a significant increase during stair descent. The patterns of stresses on the superior and inferior neck were opposite to the hip moments.

Contradict conclusions made by stresses and hip moments shows that stair ascent will generate more load on the muscles as assessed by the joint moments, but the load on the skeletal structure as assessed by the bone stress is increased during stair descent. A more comprehensive understanding can be obtained by look at both load on the muscular tissue and skeletal structure.

Based on these conclusions, researchers who want to illustrate the relationship between load mechanisms of skeletal structure and fracture-related issues should work more about stresses on skeletal structure instead of joint moments. But joint moments can be more helpful to illustrate the load mechanisms of muscular tissue for almost all physical activities.

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APPENDIX A. HIP CONTACT FORCE COMPARISONS

Table 3. Hip contact forces for stair navigation in both stair ascent and descent, forces normalized by body weight, patients here mean people with hip implant.

Reference	Subjects	Conditions	joint	Peak Magnitude
Bergmann, et al. (1995)	2 patients	Stair ascent	hip	3.5 BW
		stair descent	hip	3.7 BW
Bergmann et al. (2001)	4 patients	slow walking	Hip	2.42 BW
		normal walking	hip	2.38 BW
		fast walking	hip	2.5 BW
		stair ascent	hip	2.51 BW
Davy, et al. (1988)	1 patient	normal walking	hip	2.7 BW
		stair ascent	hip	2.6 BW
		stair descent	hip	2.7 BW
Heller et al. (2001)	4 patients	normal walking	hip	2.7 BW
		stair ascent	hip	2.65 BW
Heller et al. (2005)	1 patient	normal walking	hip	2.3 BW
		stair climbing	hip	2.35 BW
Stansfield et al. (2002)	5 patients, 5 males and 6 females	stair ascent	hip	3.1 BW
		stair descent	hip	3.8 BW
Current study (2013)	5 males and 5 females	stair ascent	hip	3.8 BW
		stair descent	hip	4.1 BW

APPENDIX B. ESTIMATED MUSCLE FORCES

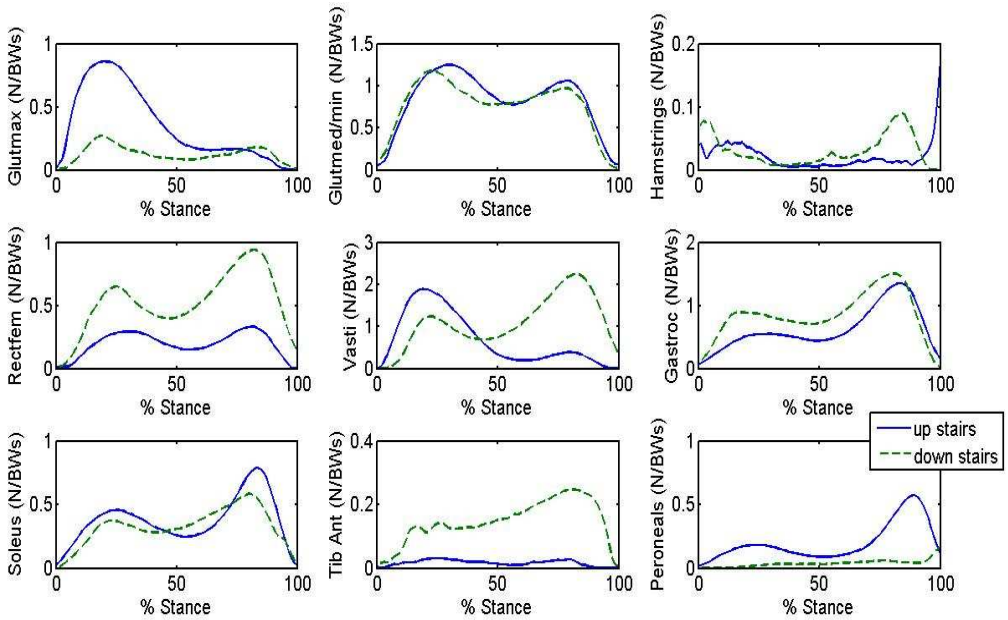

Figure 6. Ensemble average of estimated muscle forces (normalized by body weight in Newtons). Hip muscles in the 1st column, knee muscles in the 2nd column, and ankle muscles in the 3rd column. Glutmax = gluteus maximus, Glutmed/min = gluteus medius/minimus, Rectfem = rectus femoris, Vasti = vastus medialis/lateralis/intermedius, Gastroc = gastrocnemius, Tib Ant = tibialis anterior.