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Femoral Neck Stress in Older Adults During Stair Ascent and Descent

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Abstract
A detailed understanding of the hip loading environment is needed to help prevent hip fractures, minimize hip pain, rehabilitate hip injuries, and design osteogenic exercises for the hip. The purpose of this study was to compare femoral neck stress during stair ascent and descent and to identify the contribution of muscles and reaction forces to the stress environment in mature adult subjects (n = 17; age: 50–65 y). Motion analysis and inverse dynamics were combined with musculoskeletal modeling and optimization, then used as input to an elliptical femoral neck cross-sectional model to estimate femoral neck stress. Peak stress values at the 2 peaks of the bimodal stress curves (stress vs time plot) were compared between stair ascent and descent. Stair ascent had greater compressive stress than descent during the first peak at the anterior (ascent: −18.0 [7.9] MPa, descent: −12.9 [5.4] MPa, $P < .001$) and posterior (ascent: −34.4 [10.9] MPa, descent: −27.8 [10.1] MPa, $P < .001$) aspects of the femoral neck cross section. Stair descent had greater tensile stress during both peaks at the superior aspect (ascent: 1.3 [7.0] MPa, descent: 24.8 [9.7] MPa, peak 1: $P < .001$; ascent: 15.7 [6.1] MPa, descent: 18.0 [8.4] MPa, peak 2: $P = .03$) and greater compressive stress during the second peak at the inferior aspect (ascent: −43.8 [9.7] MPa, descent: −51.1 [14.3] MPa, $P = .004$). Understanding this information can provide a more comprehensive view of bone loading at the femoral neck for older population.

Keywords
bone stresses, hip contact force, estimated muscle force

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ABSTRACT

A detailed understanding of the hip loading environment is needed to help prevent hip fractures, minimize hip pain, rehabilitate hip injuries and design osteogenic exercises for the hip. The purpose of this study was to compare femoral neck stress during stair ascent and descent and to identify the contribution of muscles and reaction forces to the stress environment in mature adult subjects ($n=17$; age: 50-65 years). Motion analysis and inverse dynamics were combined with musculoskeletal modelling and optimization, then used as input to an elliptical femoral neck cross-section model to estimate femoral neck stress. Peak stress values at the two peaks of the bimodal stress curves (stress vs. time plot) were compared between stair ascent and descent. Stair ascent had greater compressive stress than descent during the first peak at the anterior (ascent: -18.0±7.9 MPa, descent: -12.9±5.4 MPa, $p<0.001$) and posterior (ascent: -34.4±10.9 MPa, descent: -27.8±10.1 MPa, $p<0.001$) aspects of the femoral neck cross section. Stair descent had greater tensile stress during both peaks at the superior aspect (ascent: 1.3±7.0 MPa, descent: 24.8±9.7 MPa, peak 1: $p<0.001$; ascent: 15.7±6.1 MPa, descent: 18.0±8.4 MPa, peak 2: $p=0.028$) and greater compressive stress during the second peak at the inferior aspect (ascent: -43.8±9.7 MPa, descent: -51.1±14.3 MPa, $p=0.004$). Understanding this information can provide a more comprehensive view of bone loading at the femoral neck for older population.

Key words: Femoral neck, bone stresses, stair ascent and descent, hip moments, estimated muscle force

Word Count: 4145

Introduction

Femoral neck fracture is a serious injury that can play an important role in morbidity and mortality among individuals, especially older adults [1]. With the overall mortality rate of hip fractures at 14.0-21.6%, the estimated 290,000 cases expected by 2030 this injury will result in a growing health problem for an aging population [2-4]. If structural failure is of concern, external loading, internal loading, bone geometry and bone material properties are the main factors that need to be investigated [5]. Therefore, it is important to investigate the loading environment of the femoral neck during activities of daily living such as stair negotiation. Modification of these activities may minimize further damage to an injured site, while still encouraging osteogenesis as a preventative measure.

A detailed analysis of the proximal femur load is necessary to understand mechanisms of failure. The femoral neck positions the hip abductor muscles away from the joint center so that adequate abductor torque can be generated to counter the large adductor torque caused by weight of the torso. In general, the torso weight vector acting on the femoral neck causes bending stress that results in inferior surface compression and superior surface tension during the single support phase of gait. Axial stress, caused by a component of the torso weight and the muscles that cross the hip, acts in compression evenly throughout the cross section and sums with the bending stress. This tends to increase the compression on the inferior surface and decrease the tension on the superior surface. The probability of bone failure may be altered through changes to the magnitude or frequency of loading (fatigue fractures) [6], insufficient bone strength (fragility fractures) [7], or a combination of these factors.

Several studies have examined the joint loading environment of the proximal femur during stair ascent and descent [8-12], yet there is still uncertainty concerning which activity produces greater loads. This lack of clarity is partly due to the difficulty in measuring the variables that are directly responsible for damage. The most direct measures of hip loading are from instrumented prostheses [8]. These devices transmit the hip joint forces via a wireless signal to a computer while patients perform various activities. It should be noted that these measurements were performed on a small number of subjects (n = 2-4) and the subjects were atypical since they had undergone hip replacement surgery within 11-31 months prior to testing. Although this procedure provides a direct measure of the hip joint forces, the invasive nature and limited subject pool reduces the practicality of this protocol in most laboratory and clinic settings.

Several studies have utilized inverse dynamics and rigid body models to estimate net joint moments and reaction forces during stair ascent and descent [13, 14]. As an estimate of femoral neck loading these measures give only a rough approximation. Both the net joint moment and the reaction force neglect the effects of co-contracting muscles and fail to consider the size of the bone in determining the potential for failure.

Using procedures developed for mechanics of materials, stress analysis is an alternative method of estimating bone loads. Inverse dynamics are combined with musculoskeletal modelling and estimated muscle forces [15] to quantify hip contact forces and ultimately stresses or strains in the bone [16]. Forces, moments and bone structure are all taken into account so that excessive loading can be determined from any source. The purpose of this study was to compare hip joint contact forces and stress on a cross section of the femoral neck during stair ascent and descent. In this study, a detailed analysis of these stresses was performed by decomposing the sources of the stress into those due to muscle forces, muscle moments, reaction forces and reaction moments. It was hypothesized that increased hip extensor muscle forces required to generate greater hip extensor moments during stair ascent [13, 14] would cause femoral neck compression stresses to be greater than during stair descent.

Methods

Seven male (age: 60 ± 6 yr; body mass: 75 ± 14 kg; height: 1.73 ± 0.05 m) and ten female subjects (age: 57 ± 5 yr; body mass: 67 ± 8 kg; height: 1.67 ± 0.05 m) who were free from lower limb injuries volunteered to participate. Before participation, they signed a written informed consent that had been approved by the Iowa State University Human Subjects Review Board.

Body mass, height, and right lower extremity segment lengths, widths, and circumferences were measured. Eighteen reflective markers were placed on anatomical landmarks of the trunk, pelvis, and right lower extremity with a minimum of 3 markers/segment: toe, heel for the foot segment; anterior/posterior leg for the leg segment; anterior thigh, right hip for the thigh segment; left hip, right/left ASIS and sacrum for the pelvis segment; medial/lateral ankle can be considered both in the foot and leg segments, medial/lateral knee can be considered both in the leg and thigh segments. All anthropometric measurements and marker placements were performed by the same researcher. A static trial was collected with the subject in anatomical position to estimate joint center locations by the markers on the joints and then medial side markers of the lower extremity were removed. All subjects performed five trials of stair ascent and five trials of descent (three-step staircase, height of each stair: 19 cm) at their normal comfortable speed. The left foot started each trial, and the right foot contacted the force platform on the second step. AMTI force platforms (1600 Hz, AMTI, Watertown, MA) were placed on the two lower stairs to measure ground reaction forces. Motion data were collected using an 8-camera system (160 Hz, Vicon MX, Centennial, CO).

Ground reaction forces and motion data were filtered using a fourth-order, low-pass Butterworth filter with a cutoff frequency of 6 Hz [17, 18]. The stance phase cycle for stair ascent/descent began when the right foot contacted the force platform and finished with toe-off. All gait cycles were normalized into a percentage of the stance phase. A rigid body model was used with inverse dynamics procedures to estimate three-dimensional joint moments and reaction forces at the ankle, knee, and hip. Segment masses, center of mass locations, and moments of inertia were obtained by the equations of Vaughan et al. [19].

An individually scaled musculoskeletal model based on the joint and muscle definitions of Arnold et al. [20] was implemented in Matlab to estimate the dynamic muscle-tendon length and velocity adjusted maximal muscle forces, muscle moment arms and orientations for 44 lower limb muscles using the three dimensional segment angles obtained during the trials. Static optimization was used to select a set of muscle forces that minimized the sum of the squared muscle stresses [21] and balanced using the sagittal plane hip, knee and ankle moments, frontal plane hip moment and the transverse plane hip and ankle moments. Solutions were also constrained by the maximal dynamic muscle forces estimated with the musculoskeletal model.

$$Min\ \sum_{i=1}^{44}(\mathrm{F_i}/\mathrm{A_i})^2 \qquad \text{Subject to: } \mathrm{r_{ij}} \times \mathrm{F_i} = \mathrm{M_j}\ \ 0 \leq \mathrm{F_i} \leq \text{Max dynamic } \mathrm{F_i}$$

For the ith muscle: F_i is estimated muscle force, A_i is the cross-sectional area, r_{ij} is the moment arm for the jth joint moment, and M_j is the jth joint moment.

Hip joint reaction forces were summed with muscle forces crossing the hip joint to obtain hip contact forces that were then transformed into the thigh coordinate system. The thigh coordinate system has the long axis of femur as longitudinal direction (y-axis), the cross product of y-axis and the vector from knee joint center to lateral knee marker as the anterior-posterior axis (x-axis), the cross product of x- and y-axis as medial-lateral axis (z-axis). Forces and moments acting at the centroid of the femoral neck cross section were calculated by transforming the hip contact forces into a femoral neck coordinate system and using the techniques and assumptions of beam theory. The femoral neck coordinate system had one axis (z-axis) parallel to the longitudinal axis of the neck and one orthogonal axis pointing approximately forward (x-axis) and the third (y-axis) obtained by the cross product of the first two.

An elliptical bone model (Figure 1) was used to estimate stresses on the superior, inferior, anterior and posterior surface of the femoral neck. Age and gender specific subperiosteal width and cortical width [22] were used to create quadratic prediction equations for the outer and inner diameters along the superior/inferior axis.

$$\text{Male Outer Diameter} = -0.0004 \times \text{age}^2 + 0.0962 \times \text{age} + 32.042, \mathrm{R}^2 = 0.982$$

$$\text{Male Inner Diameter} = -0.0004 \times \text{age}^2 + 0.1152 \times \text{age} + 27.476, \mathrm{R}^2 = 0.987$$

$$\text{Female Outer Diameter} = -0.0004 \times \text{age}^2 + 0.1036 \times \text{age} + 26.662, \mathrm{R}^2 = 0.99$$

$$\text{Female Inner Diameter} = -0.0003 \times \text{age}^2 + 0.1102 \times \text{age} + 22.445, \mathrm{R}^2 = 0.994$$

Where age is in years and diameters are in millimeters.

Anterior/posterior diameters were estimated by multiplying the superior/inferior diameters by the ratio of maximal to minimal diameters (male: 1.16±0.04; female 1.26±0.03) [23].

The stress estimation formulas were as follows:

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

$\sigma_{\text{anterior}} = \sigma(M_{\text{ml}}) + \sigma(F_{\text{axial}}) \qquad \sigma_{\text{posterior}} = \sigma(-M_{\text{ml}}) + \sigma(F_{\text{axial}})$

Where σ_{superior} is the stress on the superior aspect of the femoral neck, σ_{inferior} is the stress on the inferior aspect, σ_{anterior} is the stress on the anterior aspect, $\sigma_{\text{posterior}}$ is the stress on the posterior aspect, $\sigma(M_{\text{ml}})$ is the stress generated by sagittal plane moment, $\sigma(M_{\text{ap}})$ is the stress generated by frontal plane moment and $\sigma(F_{\text{axial}})$ is the stress caused by the axial force. Negative values indicate compressive stress and positive values indicate tensile stress.

The total stress on the femoral neck cross section is caused by a combination of the joint reaction force and the muscle forces. An analysis was undertaken to investigate how these variables independently affect the stress environment. The joint reaction force and muscle forces have the potential to compress the elliptical cross section of the femoral neck and to produce bending about the major and minor axes. The stress analysis was performed using each of these four components separately: 1) the joint reaction force compression component, 2) the muscle force compression component, 3) the joint reaction force bending/moment component, and 4) the muscle bending/moment component. The total muscle component was the sum of 2 and 4, the total reaction component was the sum of 1 and 3, and the total stress was the sum of 1-4.

Previous validation work explored the correlations between elliptical bone model and CT bone model for the tibia bone, which showed that the correlation of peak tensile stress on the anterior site was 0.89, and the peak compressive stress on the posterior site was 0.96 [24]. These correlations were considered to be sufficiently high to perform a repeated measures statistical analysis. Since the shape of the femoral neck cross section is a closer fit to the elliptical model than the tibia, it is reasonable to assume that the correlations of stresses on the femoral neck between the elliptical model and a CT model should be even higher than the outcomes from the tibia study [24]. Moreover, this elliptical model was selected since it provided a method to determine stress curves for the entire stance phase and allows for the breakdown of individual stress components in a more computationally efficient manner than with a finite element model.

The primary dependent variable was stress, but peak longitudinal, medial-lateral and anterior-posterior hip contact forces were also calculated to help explain results. Pairwise t-tests were used to compare the peak hip joint contact forces between stair ascent and descent. For the stress analysis, the independent variables were the direction of travel (ascent vs descent) and the site or location of the stress on the femoral neck cross section (anterior, posterior, superior and inferior). The stress was estimated at the two time points during the stance phase that corresponded with the two peak values on the stress by time curves. The stress on the superior aspect of the cross section did not have a consistent first peak during stair ascent so the average time of the peak during descent was used. In order to get a more holistic view of the relationship between these variables, the stress was decomposed into four sources (muscle force, muscle moment, reaction force and reaction moment). The average of 5 trials for each direction was used for statistical analysis. Positive stress values indicate a tensile stress and negative values indicate a compressive stress throughout this paper, however statistics were performed on the absolute value of the stress, making tensile and compressive stress clinically equivalent. A two-way repeated-measures MANOVA was used to compare the differences between the four sites on the femoral neck and the direction as well as a site by direction interaction (SPSS, IBM Corp). Univariate ANOVAs were performed given a significant multivariate statistic. Pairwise t-tests were used to compare the stresses at the same site between stair ascent and descent. If sphericity was violated a Greenhouse-Geisser correction was performed. Force and moment contributions to the stress were not statistically compared but used to explain stress magnitudes. The alpha level was set at .05 for all statistical tests.

Results

Hip joint contact forces acts at the center of the femoral head (hip joint center) and are presented in the thigh coordinate system (Figure 2). The anterior-posterior and longitudinal forces tended to be bimodal with the first peak occurring at approximately 20% of stance and the second peak occurring at approximately 80% of stance. In general, the peak 1 forces were greater than peak 2. The laterally directed component often had only a single peak value occurring at 20% of stance. This component of the hip joint contact force was statistically greater during ascent (ascent: 2.51 ± 0.39 BW, descent: 1.37 ± 0.24 BW, $p<0.001$). The peak 1 posteriorly directed component of the hip joint contact force was also greater in ascent (ascent: 1.44 ± 0.29 BW, descent: 0.85 ± 0.19 BW, $p<0.001$) while the peak 2 force was greater during descent (ascent: 0.51 ± 0.14 BW, descent: 0.72 ± 0.20 BW, $p=0.002$). Peak hip contact forces in the longitudinal direction had the greatest magnitudes and were directed distally. They had the same trend as the posteriorly directed force peaks – peak 1 was greater during ascent (ascent: 4.57 ± 0.53 BW, descent: 3.95 ± 0.49 BW, $p=0.001$) while peak 2 was greater during descent (ascent: 3.16 ± 0.38 BW, descent: 3.77 ± 0.63 BW, $p=0.006$).

MANOVA results revealed a significant interaction between direction (ascent vs descent) and site (anterior, posterior, superior and inferior) in peak stress ($p<0.001$) indicating a significant interaction in at least one of the peak values. MANOVA main effects were also significant for direction ($p=0.017$), and site ($p<0.001$).

Univariate results indicated that the direction by site interaction was present for both peak 1 ($p<0.001$) and peak 2 ($p=0.013$) stresses. Figure 3 and 4 illustrate these interactions by showing changes in the peak stress values between stair ascent and descent for each site. The interaction during peak 1 was due to increases in stair descent stresses on the superior and inferior aspects of the femoral neck but decreases in stair descent stresses on the anterior and posterior aspects. The interaction during peak 2 was due to a greater increase in stair descent stress on the inferior aspect of the femoral neck relative to the other three sites. Main effect statistics are not presented due to the significant interactions. Post-hoc paired t-tests were used to compare ascent vs descent total stress values at each site and peak (Table 1 and 2).

During both stress peaks the dominant loading in the femoral neck was compressive and occurred on the inferior region of the femoral neck during both ascent and descent. Compressive stress at the inferior site was greater during stair descent than ascent for the second stress peak (-43.8 ± 9.7 MPa (peak 2-Ascent-Inferior), -51.1 ± 14.3 MPa (peak 2-Descent-Inferior), $p = 0.004$).

Peak tensile stresses occurred in the superior region and were greater during stair descent during both peaks (1.3 ± 7.0 MPa (peak 1-Ascent-Superior), 24.8 ± 9.7 MPa (peak 1-Descent-Superior), $p < 0.001$; and 15.7 ± 6.1 MPa (peak 2-Ascent-Superior), 18.0 ± 8.4 MPa (peak 2-Descent-Superior), $p = 0.028$) compared to stair ascent.

The anterior and posterior regions were generally in compression. Peak compressive stress on the anterior aspect of the femoral neck was greater during stair ascent (-18.0 ± 7.9 MPa (peak 1-Ascent-Anterior)) compared to descent (-12.9 ± 5.4 MPa (peak 1-Descent-Anterior)) at peak 1 ($p < 0.001$). Likewise, the posterior aspect had an increased compressive stress for stair ascent (-34.4 ± 10.9 MPa (peak 1-Ascent-Posterior)) than descent (-27.8 ± 10.1 MPa (peak 1-Descent-Posterior)) at peak 1 ($p < 0.001$).

Based on the estimations from the model, the stress caused by the reaction component was calculated separately from the stress caused by the muscle component so that distinct contributions to the stress load could be assessed (Table 1). Overall, the greatest stresses were compressive with the reaction component causing greater stress magnitudes than the muscle component. The three greatest stress magnitudes caused by the reaction component were -86.6 ± 17.1 MPa (peak 1-Ascent-Inferior), -80.6 ± 26.2 MPa (peak 1-Descent-Inferior), and -73.0 ± 19.1 MPa (peak 2-Ascent-Inferior) while the three greatest stress magnitudes caused by the muscle component were -63.1 ± 18.5 MPa (peak 1-Ascent-Superior), -48.7 ± 17.3 MPa (peak 1-Descent-Posterior), and -43.8 ± 17.8 MPa (peak 2-Ascent-Superior).

Stresses were also decomposed according to the contributions from moments and forces. In general the contribution to the peak stress was dominated by the moments. The greatest compressive stress was at the inferior site during peak 1 of stair descent, the reaction force produced -4.1 ± 0.9 MPa of compression and the muscle forces produced -13.0 ± 3.0 MPa. However, the reaction moment produced 36.4 ± 18.1 MPa of tensile stress and the muscle moment produced -76.6 ± 25.8 MPa of compressive stress.

Discussion

The hypothesis that hip contact forces would be significantly greater during stair ascent was not universally supported by the results. Peak hip contact forces were greater during stair ascent than descent at peak 1, but at peak 2 the posteriorly and distally directed hip contact forces were greater during stair descent than ascent (Figure 2). These shapes of the contact force curves were mirrored by the muscle activity in the hip extensor muscles during ascent and descent (Figures 6 and 7).

The hypothesis that femoral neck stress would be significantly greater during stair ascent was also not supported by the results. We estimated femoral neck stresses at four sites on the femoral neck during stair ascent and descent for older adults and then analyzed the sources of stress. The MANOVA main effect of ascent/descent on femoral neck stresses ($p = 0.017$) and the interaction effect between directions and femoral neck sites ($p < 0.001$) were both significant.

The univariate interaction effects for both stress peak 1 ($p<0.001$) and peak 2 ($p=0.013$) were significant. Results indicates that 1) at some sites the stresses were greater during ascent than descent, in other sites the stresses were greater during descent compared to ascent (Figure 3); 2) stress change patterns were similar among different sites, but the change of slope between stair ascent and descent for some sites were much greater than other sites (Figure 4). Both peak tensile stress at the superior site (both peaks) and peak compressive stress at the inferior site (peak 2) showed greater stress during stair descent. The peak 1 stress during early stair descent could be a consequence of a relatively extended position of the hip and knee during this phase of the decent. This erect posture may allow the ground reaction force vector to be directed through the joints and minimize the ability of the muscles to absorb the energy of the downward moving mass. This can be seen in the EMG activity of the hip extensor muscles during stair ascent and descent [25].

An examination of the stress caused by the reaction force/moment compared to the muscle force/moment highlights how this relationship affects the total stress environment. In general, any time the reaction force/moment caused a stress greater than 25 MPa (compressive or tensile) the muscles contracted to produce bending in the opposite direction and thus reduced the total stress. For example, the weight and acceleration of the torso caused a reaction force to push down on the head of the femur during P1 ascent. This bent the neck concave inferior and caused 64.4 ± 17.1MPa of tensile stress on the superior aspect of the neck. However, hip abductor and extensor muscles were activated at that time and they produce concave superior bending. This compressed the superior region with a stress of -63.1 ± 18.5 MPa. The net result was minimal stress (1.3 ± 7.0 MPa) because the tensile stress cancelled the compressive stress in this region. During descent the reaction component caused 67.5 ± 24.7 MPa of tensile stress while the muscle component was reduced to -42.7 ± 19.0 MPa of compression. This resulted in increased tension on the superior surface (24.8 ± 9.7 MPa) compared to ascent (1.3 ± 7.0 MPa).

On both superior and inferior surfaces of femoral neck, the muscle component produced stresses opposite to, but smaller in magnitude than the stresses produced by the reaction force/moment, so greater stresses from muscle can be an effective way to minimize the net stresses on these 2 surfaces of femoral neck. Stair descent tended to decrease the stresses produced by muscle compared to stair ascent. This suggests that the ability of the muscles to reduce bone stress may be minimized during stair descent.

The stress produced by forces is predominantly compressive, while stress produced by moments creates compression on the inside of the curvature and tension on the outside portion. In general, the magnitude of the stress caused by moments was greater than the magnitude caused by forces at most sites and directions. The stress caused by the moment dominated on both the inferior and superior regions but the contribution of the moments was generally reduced at the anterior and posterior sites.

There are several limitations associated with these procedures. An ellipse model of the femoral neck cross-section was created for each subject based on age and gender. Derrick et al. [24] showed that a homogeneous elliptical model such as this could be favorably compared to a more detailed nonhomogeneous model derived from a CT scan of the tibial cross-section (r-squared = 0.89 for the peak tensile stress on the anterior site, and 0.96 for the peak compressive stress on the posterior site). Moreover, the shape of femoral neck cross section is more elliptical than the cross section of the tibia, suggesting these correlations are conservative when compared to the current study. Muscles were scaled to the individual but modeled with standardized insertions, origins and contraction properties and the muscle forces were estimated using static optimization with a cost function that minimized muscle stresses. Individual differences in the muscle properties or non-optimal sequencing of muscle activity could influence the muscle forces.

Muscle optimization does not guarantee an exact replication of the muscle force patterns. The process assumes that the activation of muscles follows the rules of the cost function (minimization of the sum of the muscle stress squared). Differences between the estimated and actual muscle force may occur if a person uses an alternate pattern of recruitment. Estimated hip extensor muscle forces were compared to EMG data [25] in the literature to assess the accuracy of this estimation. Figures 6 and 7 show the comparison between hip extensor muscle forces and EMG activity (including biceps femoris long head, semimembranosus, upper gluteus maximus, and gluteus medius muscle forces). The cross-correlation between the two EMG and muscle force shows acceptable agreement for stair ascent (0.725) but a lower cross-correlation value (0.162) for descent. This lower cross-correlation during descent was due to the peaks of EMG curves being shifted closer to contact and toe-off.

The lower limb joint center (ankle/knee/hip) calculations were based on the reflective markers placed on the bony landmarks of each joint. Inaccuracies in marker placement may result in errors in the estimation of joint center locations. These inaccuracies may decrease the accuracy of joint moment estimation based on the inverse dynamics method [26]. Although the repeated measures nature of this study likely reduces the effect of inaccurate marker placement it does not insure that the errors in marker placement or marker movement are the same between ascent and descent.

In this study, contact force and femoral neck stress were analyzed to evaluate loading at the hip joint during stair ascent and descent for older population. Joint contact forces give a good estimate of the loading between the femoral head and the pelvic acetabulum. Bone stresses are more directly related to the loads that cause the bone to fracture and include the influence of muscle forces, reaction forces and bone geometry. We found that the greatest hip contact forces occurred at about 20% of stance while ascending the stairs. Stresses in the femoral neck were generally, but not universally, greater during stair descent. Stress variations on the periphery of the femoral neck cross-section were large with the inferior region receiving the greatest stress values. Combining contact forces and bone stresses could help future studies analyze loading conditions in a more comprehensive way for other physical activities.

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Figures

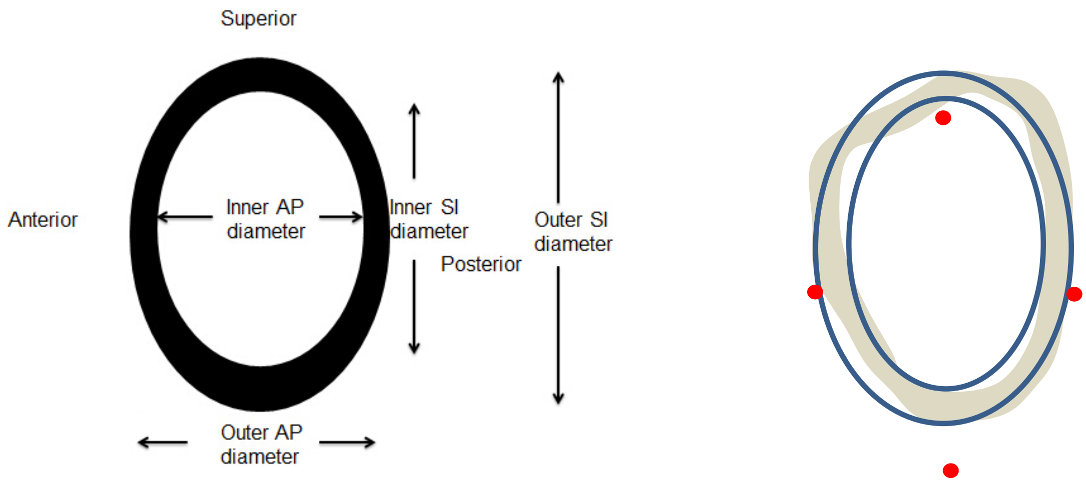

Figure 1. Elliptical bone model superimposed on a cross-sectional CT scan of the femoral neck.

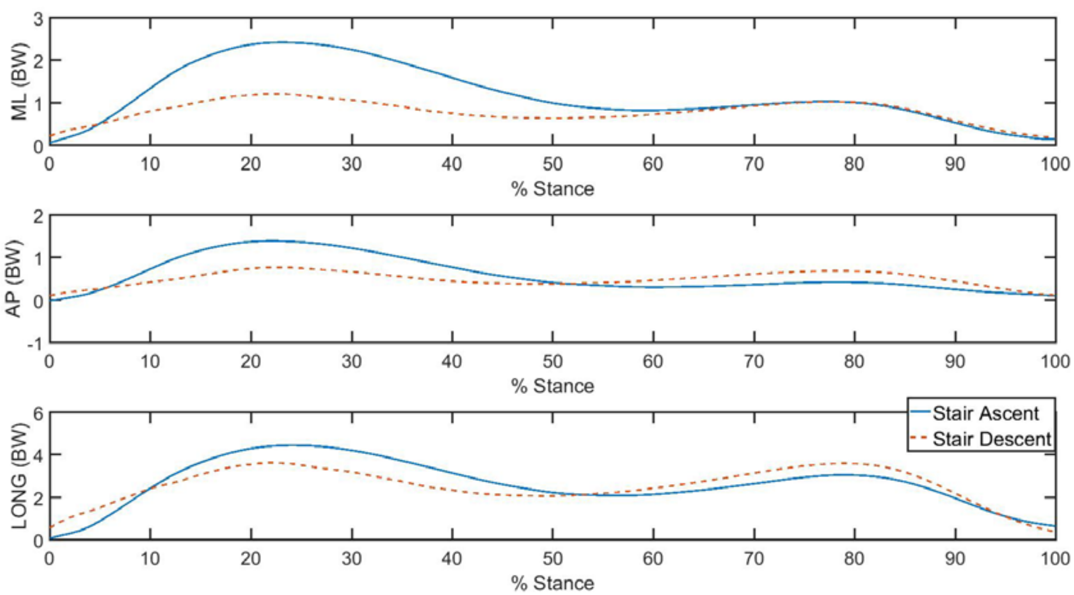

Figure 2. Ensemble average of contact forces at 3 planes of the hip joint. ML stands for Medial-Lateral direction, AP stands for Anterior-Posterior direction, LONG stands for longitudinal direction. Positive values indicate lateral, posterior, and downward directions.

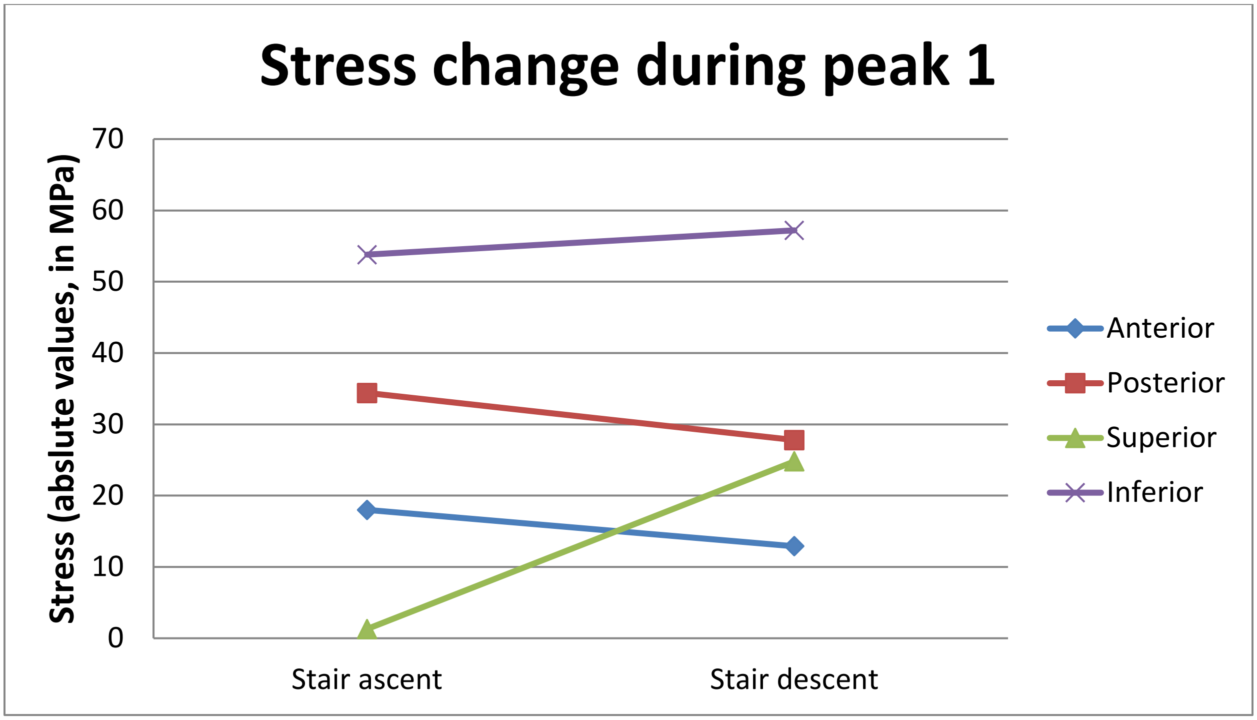

Figure 3. Change of stress between stair ascent and descent during stress peak 1 for the anterior, posterior, superior and inferior regions of the femoral neck.

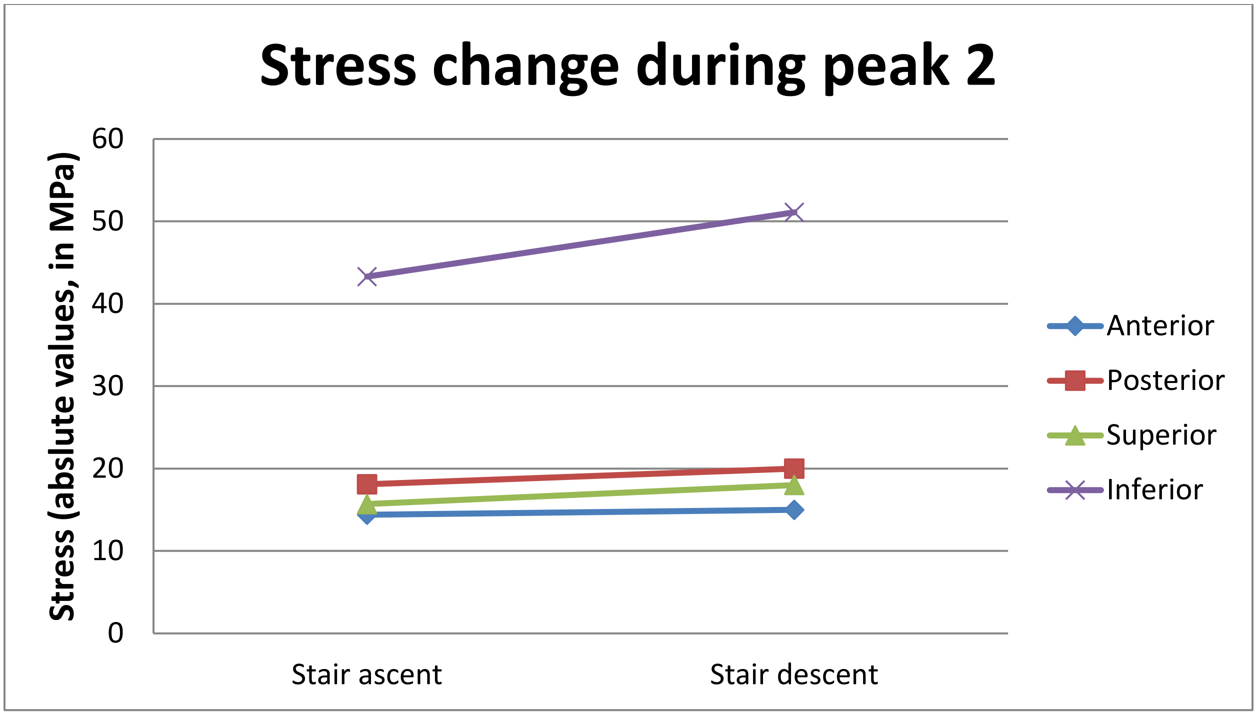

Figure 4. Change of stress between stair ascent and descent during stress peak 2 for the anterior, posterior, superior and inferior regions of the femoral neck.

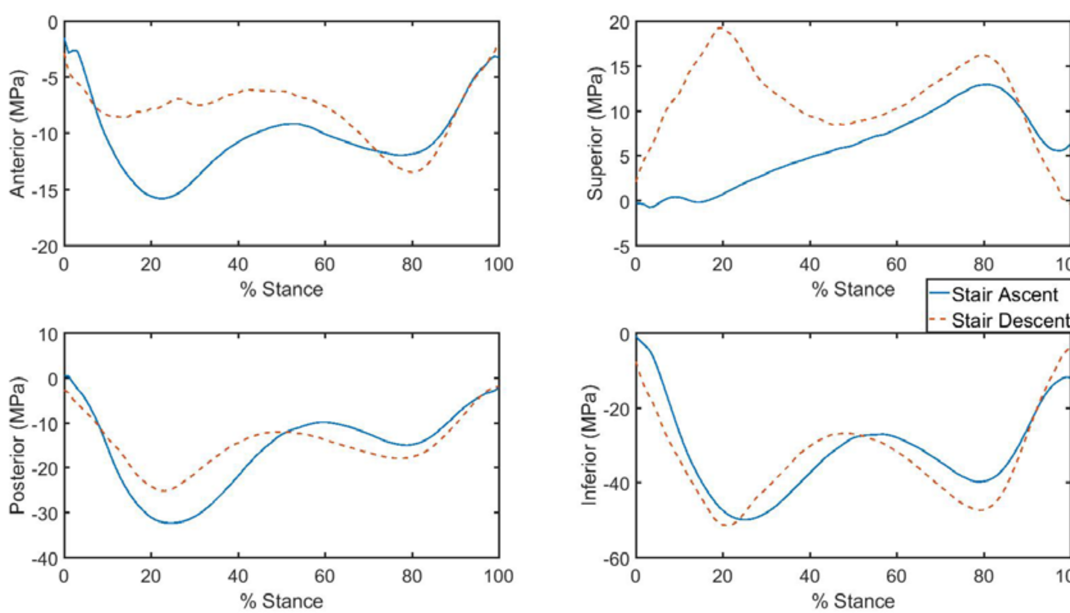

Figure 5. Ensemble average of stresses at superior and inferior sites of femoral neck, positive values indicate tension, negative indicate compression.

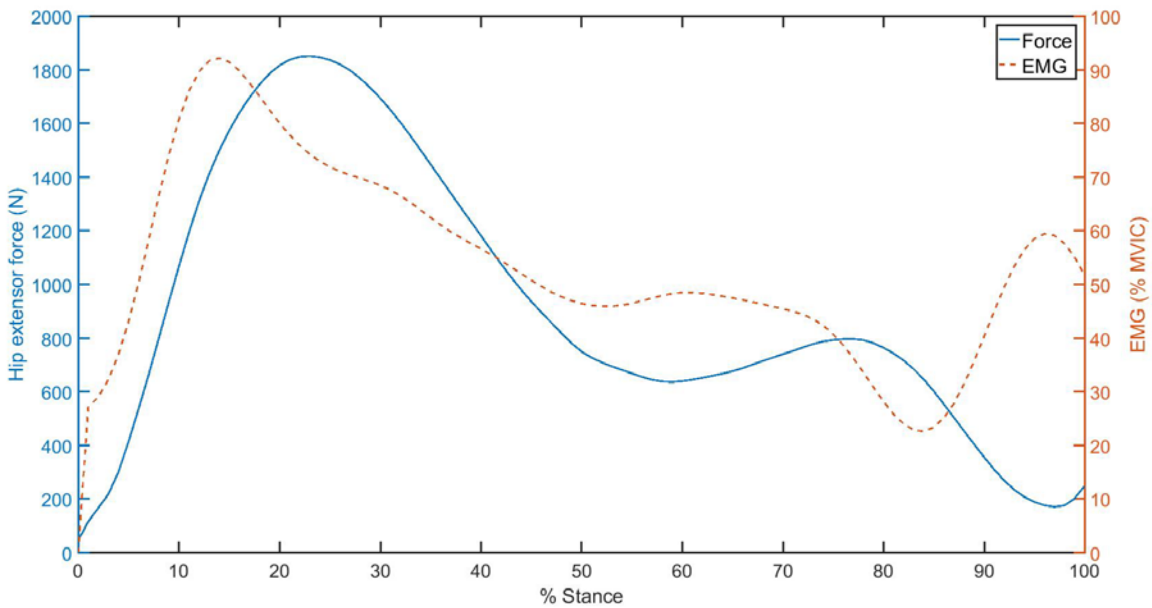

Figure 6. Average of estimated hip extensor muscle forces (in Newtons) and EMG activities (in % MVIC) [19] during stair ascent.

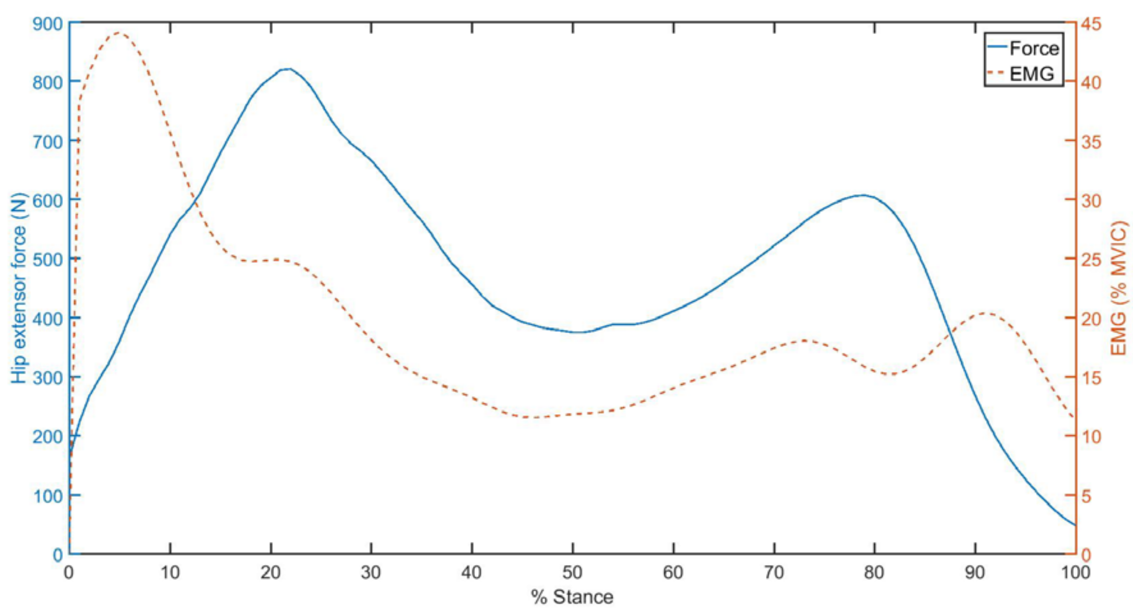

Figure 7. Average of estimated hip extensor muscle forces (in Newtons) and EMG activities (in % MVIC) [19] during stair descent.

Tables

Table 1. Means (SD) of muscle caused and reaction force caused stresses (MPa) on 4 sites on the femoral neck during stair ascent and descent for peak 1. Bolded values and corresponding p-values indicate significant differences between stair ascent and descent stresses.

Stress Site	Peak 1 Stress Components													
	Stair Ascent Stress (MPa)							Stair Descent Stress (MPa)						
	Muscle			Reaction				Muscle			Reaction			
	Force Source	Moment Source	Total Muscle	Force Source	Moment Source	Total Reaction	**Total**	Force Source	Moment Source	Total Muscle	Force Source	Moment Source	Total Reaction	**Total**
Superior	-20.6	-42.6	-63.1	-2.9	67.3	64.4	**1.3**[1]	-10.7	-31.9	-42.7	-3.8	71.3	67.5	**24.8**[1]
	(4.5)	(15.7)	(18.5)	(0.6)	(18.5)	(17.1)	(7.0)	(3.2)	(16.0)	(19.0)	(1.0)	(24.0)	(24.7)	(9.7)
Inferior	-21.7	54.5	32.8	-3.4	-83.2	-86.6	-53.8	-13.0	36.4	23.5	-4.1	-76.6	-80.6	-57.2
	(4.9)	(20.6)	(17.2)	(0.6)	(22.9)	(23.4)	(12.2)	(3.0)	(18.1)	(16.4)	(0.9)	(25.8)	(26.2)	(15.1)
Anterior	-20.0	28.0	8.0	-3.0	-23.0	-26.0	**-18.0**[1]	-12.3	-2.3	-14.6	-3.1	4.9	1.8	**-12.9**[1]
	(6.2)	(12.2)	(9.0)	(1.1)	(12.1)	(12.7)	(7.9)	(2.6)	(12.5)	(11.0)	(1.1)	(13.9)	(14.4)	(5.4)
Posterior	-21.9	-26.8	-48.7	-3.4	17.7	14.3	**-34.4**[1]	-13.4	7.1	-6.3	-4.0	-17.6	-21.6	**-27.8**[1]
	(4.9)	(13.4)	(17.3)	(0.6)	(16.0)	(15.8)	(10.9)	(3.0)	(12.4)	(13.2)	(0.8)	(15.6)	(15.8)	(10.1)

1. $p < 0.001$

Table 2. Means (SD) of muscle caused and reaction force caused stresses (MPa) on 4 sites on the femoral neck during stair ascent and descent for peak 2. Bolded values and corresponding p-values indicate significant differences between stair ascent and descent stresses.

Stress Site	**Peak 2 Stress Components**													
	Stair Ascent Stress (MPa)							Stair Descent Stress (MPa)						
	Muscle			Reaction				Muscle			Reaction			
	Force Source	Moment Source	Total Muscle	Force Source	Moment Source	Total Reaction	**Total**	Force Source	Moment Source	Total Muscle	Force Source	Moment Source	Total Reaction	**Total**
Superior	-9.9 (2.1)	-33.9 (16.4)	-43.8 (17.8)	-3.0 (0.8)	62.6 (20.0)	59.5 (19.3)	**15.7**[1] (6.1)	-13.2 (3.4)	-15.2 (10.5)	-28.4 (12.1)	-2.7 (0.6)	49.2 (18.6)	46.5 (18.2)	**18.0**[1] (8.4)
Inferior	-10.9 (2.4)	40.1 (15.8)	29.2 (14.3)	-3.4 (0.6)	-69.6 (18.6)	-73.0 (19.1)	**-43.8**[2] (9.7)	-13.9 (3.4)	16.1 (11.3)	2.3 (10.3)	-2.8 (0.6)	-50.6 (19.3)	-53.3 (19.9)	**-51.1**[2] (14.3)
Anterior	-9.2 (2.8)	5.0 (9.5)	-4.2 (8.7)	-2.8 (0.8)	-7.4 (11.1)	-10.2 (11.6)	-14.4 (7.7)	-13.2 (2.8)	-30.2 (11.8)	-43.6 (13.2)	-2.6 (0.6)	31.0 (11.6)	28.5 (11.4)	-15.0 (5.3)
Posterior	-10.6 (2.3)	-6.3 (9.7)	-16.9 (10.3)	-3.0 (0.7)	1.8 (11.3)	-1.2 (11.3)	-18.1 (8.6)	-13.2 (3.3)	30.0 (12.5)	16.7 (11.4)	-2.7 (0.8)	-34.0 (14.5)	-36.7 (14.8)	-20.0 (7.6)

1. $p < 0.029$
2. $p < 0.005$