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The Influence of Impact Angle on Hip Fracture in Fallings

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Abstract : The direction of the applied load and displacement rate on the femur during falls may be an important factors in the etiology of hip fractures. Nonetheless, previous studies did not consider these two factors simultaneously for falling condition. Therefore, in the present study, an impact test system is developed to simulate the falling condition and the influence of impact angle on the deformation pattern changes of proximal femur is investigated. The results showed that a slight variation in impact angle quite affects deformation pattern of the proximal femur. Along with bone mineral density and trabaecular morphology, the impact angle can be another important factor affecting the structural capacity of the proximal femur.

Key words: femur, falls, impact test, hip fracture

1. Introduction

Falling related injuries are categorized as the most serious and common medical problems experienced by the elderly. Hip joint fracture is one of the most serious consequences of falling in the elderly, and accounts for a considerable part of the disability, death, and medical costs associated with falling. Since fracture represents a structural failure of bone, I believe that such reductions in fracture incidence can only be achieved through a sound understanding of the biomechanics of fracture etiology [1]. Early research on hip fracture biomechanics focused primarily on loads associated with gait or one-legged stance [2]. However, since spontaneous fractures are rare and approximately 90% of hip fractures are associated with a fall [3], the mechanics of the descent and impact phases of a fall are of major etiologic importance [4]. Although a fall appears to be a necessary condition for hip fracture, it is not a sufficient condition [5], since less than 5% of falls results in fractures [6]. Additional importance elements in the etiology of hip fracture are the severity of the fall and the structural capacity of the femur.

The available energy, impact location, and muscle activity during descent affect the severity of a fall, whereas bone density, architecture, and the geometry influence the structural capacity of the hip [7].

Femoral bone density (BMD) has proven to be an effective predictor of fracture risk. By predicting a fracture risk from an individual's BMD, one can estimate the denominator of the factor of risk. Other factors, which are independent of density, might also affect the structural capacity of the femur. For example, Courtney *et al.* [8] reported a 20% increase in failure load due to high displacement rates. Testing at a lower displacement rate, Lotz and Hayes [9] reported a lower average load than Courtney *et al.* [10] for elderly femurs. However, the direction of the applied load from Lotz et al. differed from that of Courtney *et al.* Thus, while density appears to be an important determinant of fracture load, loading angle and displacement rate may also be crucial elements [1].

The direction of the applied load and displacement rate on the femur may be an important factors in the etiology of hip fractures. Nonetheless, in previous studies, there have not been attempts to match these two factors to falling condition.

Therefore, primary aims in the present study are (1) to develop the impact testing system to simulate the falling condition; (2) to investigate the deformation pattern changes of proximal femur to consider the influence of impact angle; and (3) to conduct the traditional

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static test simulating two-legged stance to compare with deformation pattern caused by the impact test.

2. Materials and Methods

Fresh right femur (male, age 31) was stored at -70°C and thawed to room temperature prior to testing. To prevent drying during thawing, preparation, and testing, femur was periodically sprayed with distilled water.

In preparation for impact testing, the femur was cut approximately 15 cm away from the midgreater trochanter to the distal end. A 7 cm long and 1 cm diameter round rod, fixed to a steel base connected to the pelvic components of the surrogate-pelvis, was inserted in medullary canal and cemented in place with polymethylmethacrylate, and then was instrumented with 8 strain gage rosettes. The location of the gages was such that three rosettes were coplaner at specified locations: subcapital and basicervical, and two rosettes were at subtrochanteric. For measuring the strain in axial direction and transverse direction, the gages stacked, 90-degree T rosettes with an active gage length of 3.18 mm (AE-11-TS50N-120-EL, CAS) were used. The axial direction was determined along the long axis of the femoral neck. The transverse direction was done along the normal to the axial direction. The bone surface was prepared and gages bonded using the protocol of Carter *et al.* [11] (Fig. 1, Fig. 2).

After gage preparation, the femur was imaged using CT scanner (GE 9800, GE Corp). The scans were made at the site of strain gaging in subcapital and basicervical region for cross sectional geometry and gage locations for analyses of geometric properties at the gaged section. This enabled computation of the distribution of strains normal to the bone's cross sectional plane, which was useful for comparing strain patterns in each section. The

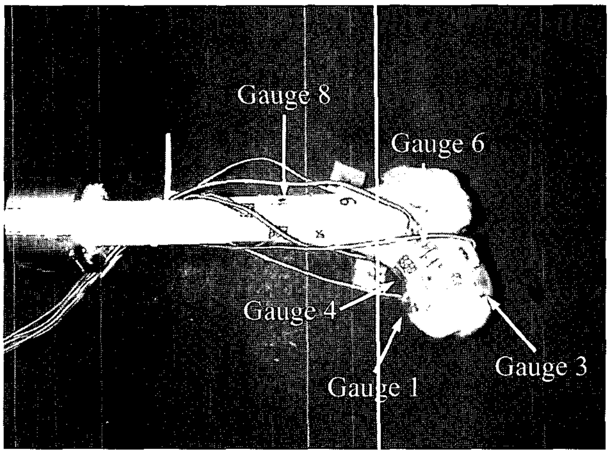

Fig. 1. Anterior view of right femur.

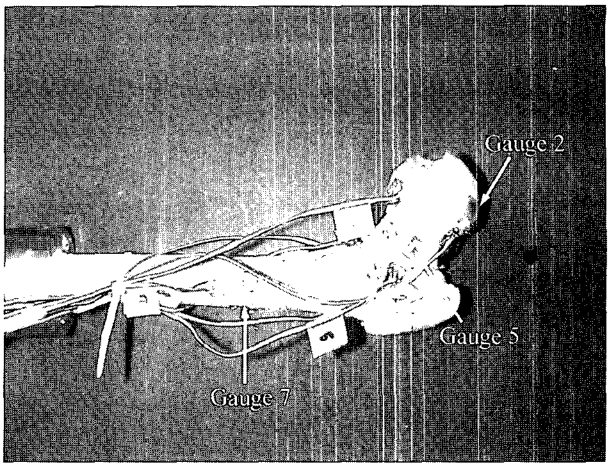

Fig. 2. Posterior view of right femur.

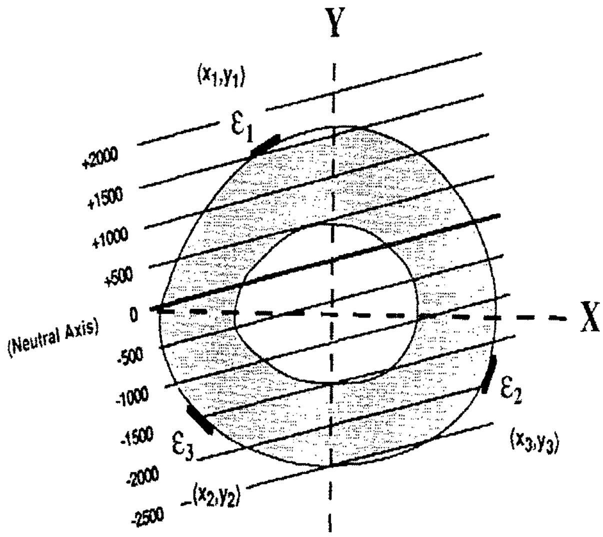

Fig. 3. A drawing of a bone cross section showing the distribution of normal strains.

computation was accomplished by a simple matrix operation on the measured strains based on strain gage location [12]. (Fig. 3).

$$\begin{bmatrix} \varepsilon_1 \\ \varepsilon_2 \\ \varepsilon_3 \end{bmatrix} = \begin{bmatrix} x_1 & y_1 & c \\ x_2 & y_2 & c \\ x_3 & y_c & c \end{bmatrix} \begin{bmatrix} a \\ b \\ c \end{bmatrix}$$

And then each gage was connected signal conditioning amplifier (model 2311, Measurement Group). The data were collected from A/D converter (DT7102, Data Translation) using a personal computer and commercial data aquisition software (DT VEE, Data Translation).

To simulate the real falling condition, the impact system is constructed, which is composed of surrogate-pelvis, impact tester, and data aquisition modules. This test system was validated with matching its effective stiffness and effective damping to the results of the pelvis-release test conducted by Robinovitch *et al.* [13].

For impact test, I set the angle of the pendulum to 13°, which yields an impact velocity of about 1 m/s, below the mean impact velocity of the hip during falling [14]. The object of this study is to investigate the relationship between impact angle and deformation pattern. Therefore, it needed to get the data from the same location at each impact angle and the test was conducted on the same specimen within the elastic range to prevent its permanent deformation. The femur was positioned so as to simulate a fall on the greater trochanter at three different impacting positions. The 15° loading configuration was identical to that used by Courtney *et al.* [10] and was typical body configuration at impact. The angle of the femoral shaft with respect to the horizontal (defined as femoral shaft angle) was 10° and in this case the femoral neck was internally rotated (defined as loading angle) by 15°. For the second loading case, the impact angle was 0° simulating impact on the hip rolled slightly forward. For the third case, the femur was internally rotated 30°.

The femoral angle with respect to the vertical was kept at 10° for all three cases. Local crushing of the greater trochanter was prevented by soft tissue padding. And a polyethylene acetabulum was also inserted between the femoral head and pelvic component.

During impact, the acceleration of the pendulum head was measured using an accelerometer (CXL04M1, Crossbow) mounted on the pendulum weight. The impact force was determined by the product of acceleration and effective mass of the pendulum. Data were acquired at a rate of 4 kHz for 500 ms. To eliminate high frequency transients in the force signal observed at the instant of impact. the data were filtered using a second-order low-pass Butterworth filter with cut-off frequency of 50 Hz (Matlab, The Math Works). And in order to eliminate potential phase change effects of the filtering, data were first filtered in the forward direction and then were run back through the filter in the reverse direction. This filtering process was observed to have little effect on either the magnitude of peak force or the instance of peak force.

After impact test, the static test was conducted to compare strain patterns in each case. In order to approximate the loading condition in two-legged stance, the anteversion was set to 14° and the ideal geometric angle, which is angle between the ideal axis (longitudinal) and the axis through the center of the femoral head from the inter-condylar notch, was set to 10°, and distal end of femur was rigidly fixed. Using a hydraulic material testing machine (Instron 8511, Instron Corp), a compressive load up to 245 N was applied to the superior aspect of the femoral head. The data were recorded at a sampling rate of 10 Hz for 2 min.

3. Result and Discussions

In this study, fall impact simulation was conducted to determine how impact angle influences the deformation pattern in proximal femur. The gage locations applied in femoral head and neck are presented with their abbreviations in Fig. 4 and Fig. 5, respectively. The axial and transverse strain patterns at the corresponding locations are presented with impact angles in Fig. 6 and Fig. 7, respectively. The maximum impact force was 1051±35N and the maximum velocity was 0.91±0.04 m/s. This maximum force is far below the mean fracture load (7200±1090N). After testing, any defects in specimen were not observed. From these points, it is proposed that the deformed

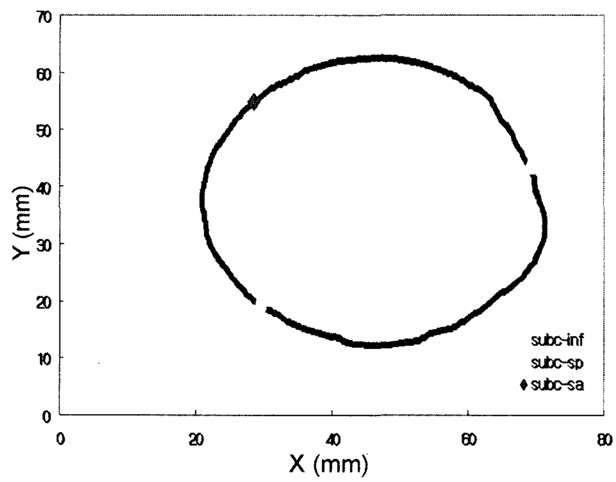

Fig. 4. Gage locations in femoral head.

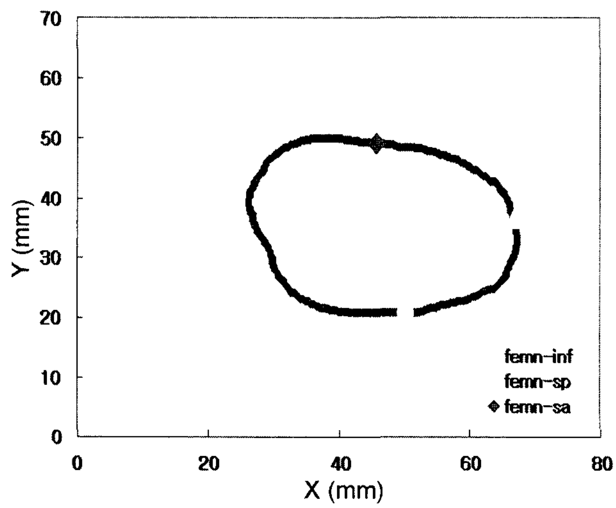

Fig. 5. Gage locations in femoral neck.

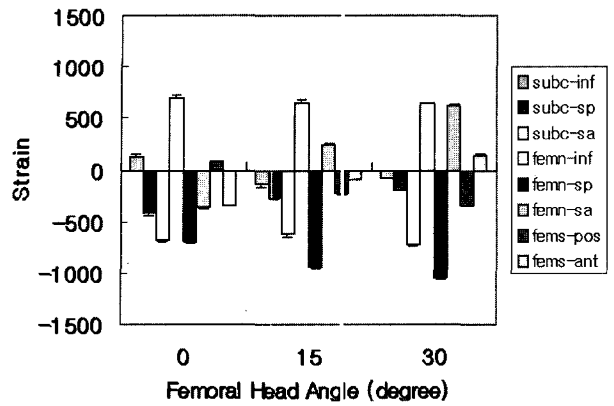

Fig. 6. Axial strain patterns at each impact angle.

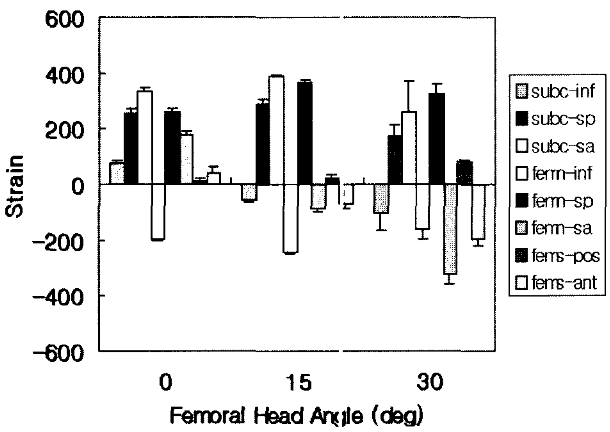

Fig. 7. Transverse strain patterns at each impact angle.

state of the specimen is still within the elastic region and the specimen doesn't experience any local permanent deformation.

The neutral axis to horizontal line in femoral head is 66.04°, 53.04°, 57.43° for impact angle 0°, 15°, 30°, respectively (Fig. 8). In femoral neck, it is -70.88°, -47.43°,

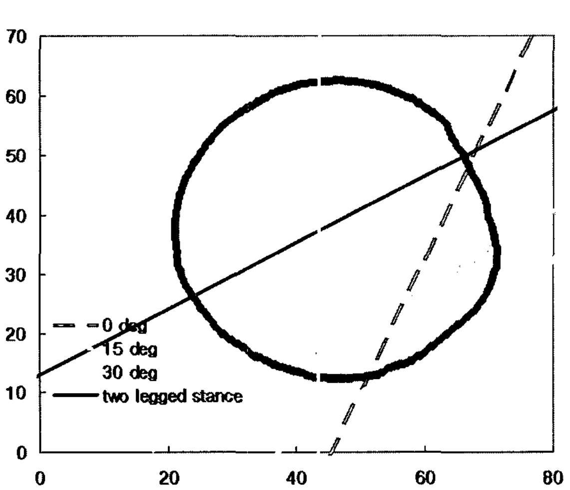

Fig. 8. Changes in neutral axis at femoral head.

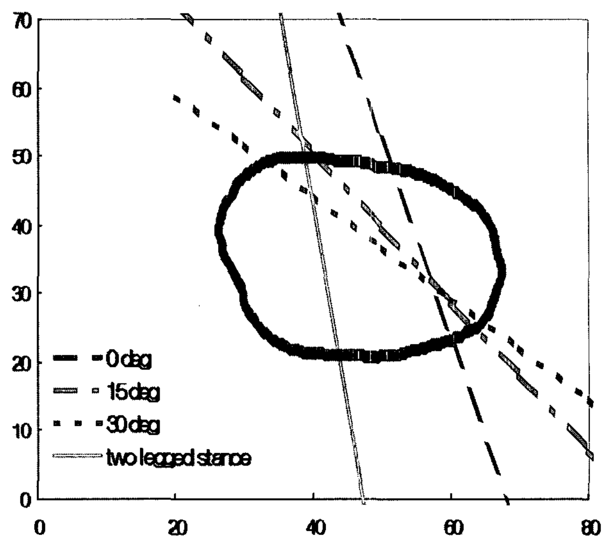

Fig. 9. Changes in neutral axis at femoral neck.

-36.46° for impact angle 0°, 15°, 30°, respectively (Fig. 9). These data indicate that the strain distribution of proximal femur is sensitive to variations in impact angle which might occur from different falls on the hip. This phenomenon is concerned with variations in moment arm between the loading point and the mid-neck. With a longer moment arm, the same impact force produces more deformation in the femoral neck. Regardless of impact angle, a larger deformation (-2000~1000 $\mu\varepsilon$) was found in femoral neck compared to femoral head (-800~200 $\mu\varepsilon$) as shown in Fig. 10 and Fig. 11. From this finding, it is suggested that the neck fracture may be one of the most typical hip fractures.

A significant difference in strain distribution and magnitude between impact test and static test was found. The neutral axis in static test was 29.05° in femoral head (Fig. 8), and -80.36° in femoral neck (Fig. 9). As shown in Fig. 12 and Fig. 13, the extend of strain distribution was -200~200 $\mu\varepsilon$ in femoral head and -300~200 $\mu\varepsilon$ in femoral neck. These results indicate that a off-axis loading was occurred in proximal femur by fall, that is, bone matrix (trabaecular bone) is so loaded obliquely relative to primary loading orientation that it fails at less strain level. Therefore, the present findings may provide another explanation for the reason why the hip fracture occurs more frequently during fall than in normal physiological loading conditions such as walking, standing, and so on.

One of the characteristics of this study is that it is the impact test to investigate specifically the effect of impact direction on the deformation of the proximal femur during fall. Unlike previous mechanical test protocols [8, 9, 10, 15], the impact test system composed of impact pendulum and surrogate pelvis was developed in the

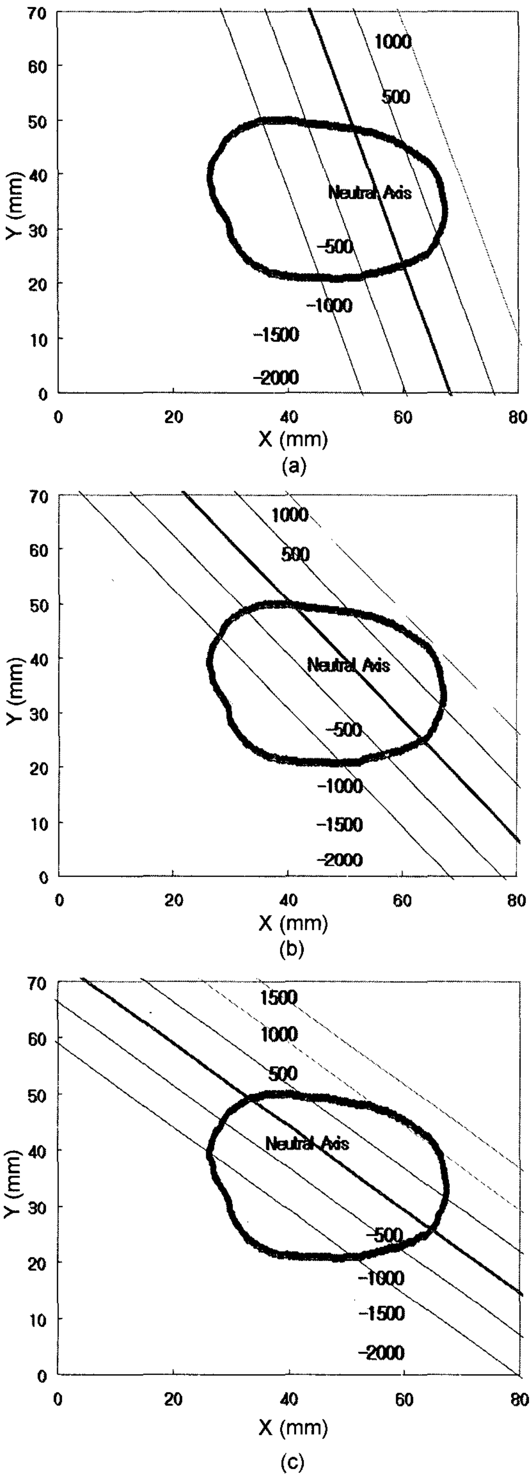

Fig. 10. Strain gradient at femoral neck for impact test: (a) impact angle 0°, (b) impact angle 15°, (c) impact angle 30°.

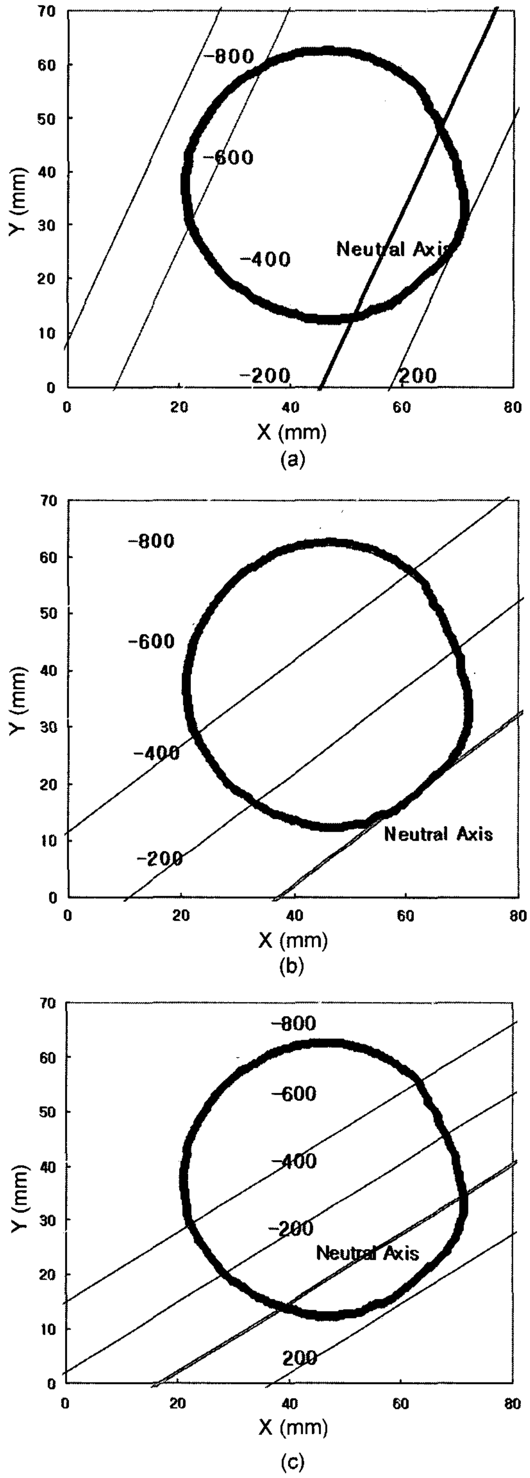

Fig. 11. Strain gradient at femoral head for impact test: (a) impact angle 0°, (b) impact angle 15°, (c) impact angle 30°.

present study. It is an essential aspect that the test system accurately simulates the dynamic response of the body during a fall on the hip. So, I adjusted the test system to simulate effective mass, stiffness, and damping of a typical male pelvis during a fall on the hip. The result well agreed with the previous study [16], which showed that impact direction affects a failure load of the femur tested in vitro. This investigation is reasonable and further emphasizes the importance of fall biomechanics in hip fracture. Another feature of the

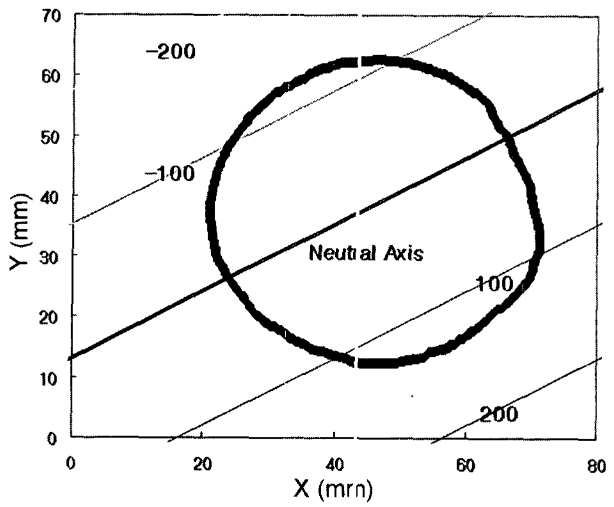

Fig. 12. Strain gradient at femoral head for static test: two-legged stance.

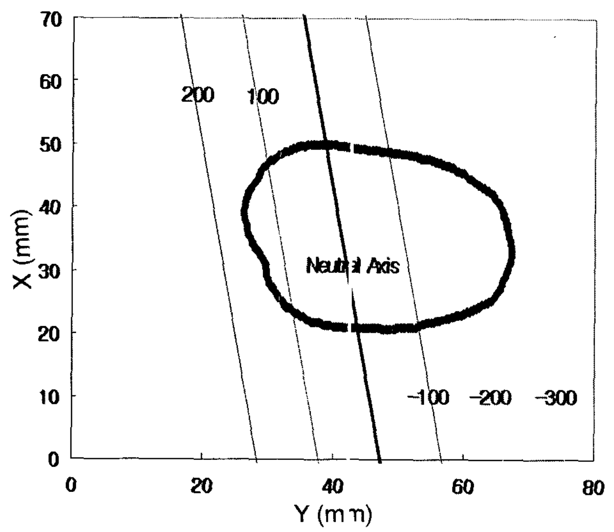

Fig. 13. Strain gradient at femoral neck for static test : two-legged stance.

present study is that, based on these data, more accurate finite element model will be constructed to predict the failure of femur. Because of few experimental data to validate models, earlier computational models have not been developed enough to predict hip fracture. So, the present findings will be very helpful. Finally, the information for the deformation patterns computed in this study is clinically useful. The computed strains are considered to be more useful in comparison between injury occurrences and biomechanical responses. Furthermore, fundamental mechanics of induced compression and bending due to a primarily impact can be analyzed in regard to the various modes of femoral fracture.

This study also has some limitations. One involves the assumption that the large local deformation is concerned with failure of bone. Load-bearing is one of the most important functions in skeletal system. In this repect, trabaecular bone may particularly play an important role in femur. Trabaecular bone failure can be fundamentally explained as a function of the strain that bone experiences. Therefore, the strain in the cross section was computed to investigate the relationship between impact angle and deformation pattern. However, the test was conducted within elastic range so that it could not be verified that bone failure started in region where the largest deformation occurred. More studies for the relationship between fracture pattern and strain distribution are needed. Secondly, there may be other impact angles that occur during falls, and perhaps the deformation pattern would be different at other angles. A better method to assess the appropriate loading condition in vivo may be to use magnetic resonance imaging or computed tomographical imaging for the hip in a fall configuration and analyze the images to estimate the angel at impact. A third limitation is the fact that there are three independent strain components about the principal centroidal axes (compression-elongation, anterior-posterior, lateral-medial responses). These components are useful for biomechanical studies of bone injury. Because of the porosity, principal axis and moment of inertia in cross section were not determined in this study.

4. Conclusions

The impact test system simulating a fall on the hip was developed in the present study. It was shown that a slight variation in impact angle affects deformation pattern of the proximal femur. Along with bone mineral density, geometry, and trabaecular morphology, the impact angle is also another important factor affecting the structural capacity of the proximal femur. Since the impact angle is mere the consequence of the direction of the fall unlike the intrinsic property of bones, the present study provides additional informations for the etiology of hip fracture in fall biomechanics.

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References

[1] T. P. Pinilla, K. C. Boardman, M. L. Bouxsein, E. R. Myers, W. C. Hayes, "Impact direction from a fall influences the failure load of the proximal femur as much as

age-related bone loss", Calicif. Tissue Int., Vol. 58, pp. 231-235, 1996.

[2] T. J. Beck, C. B. Ruff, K. E. Warden, W. W. Scott, G. U. Rao, "Predicting femoral neck strength from bone mineral data", Invest. Radiol., Vol. 25, No. 1, pp. 6-18, 1990.

[3] J. D. Michelson, A. Myers, R. Jinnah, Q. Cox, M. V. Natta, "Epidemiology of hip fractures among the elderly", Clin. Orthop., Vol. 311, pp. 129-135, 1995.

[4] S. R. Cumming, M. C. Nevitt, "Non-skeletal determinants of fractures: the potential importance of the mechanics of falls", Osteoporosis Int. (suppl), Vol. 1, pp. 67-70, 1994.

[5] S. R. Cumming, M. C. Nevitt, "A hypothesis: the cause of hip fractues", J. Gerontology, Vol. 44, No. 4, pp. 107-111, 1989.

[6] S. L. Greenspan, E. R. Myers, L. A. Maitland, N. M. Resnick, W. C. Hayes, "Fall severity and bone mineral density as risk factors for hip fracture in ambulatory elderly", JAMA, Vol. 271, pp. 128-133, 1994.

[7] E. R. Myers, W. C. Hayes, "Age-related hip fractures", Curr. Opin. Orthop., Vol. 5, pp. 9-15, 1994.

[8] A. C. Courtney, E. F. Watchel, E. R. Myers, W. C. Hayes, "Effects of loading rate on the strength of the proximal femur", Calcif. Tissue Int., Vol. 55, pp. 53-58, 1994.

[9] J. C. Lotz, W. C. Hayes, "The use of quantitative computed tomography to estimate risk of fracture of the hip from falls", J. Bone Joint Surg. [Am], Vol. 72, No. 5, pp. 689-700, 1990.

[10] A. C. Courtney, E. F. Wachtel, E. R. Myers, W. C. Hayes, "Age-related reductions in the strength of the femur tested in a fall-loading configuration", J. Bone Joint Surg. [Am], Vol. 77, pp. 387-395, 1995.

[11] J. B. Finay, R. B. Bourne, J. McLean, "A technique for the *in vitro* measurement of principal strains in the human tibia", J. Biomech. Vol. 15, pp. 723-739, 1982.

[12] A. A. Biewener, Biomechanics - Structures and Systems, IRL PRESS, pp. 129-130, 1992.

[13] S. N. Robinovitch, T. A. McMahon, W. C. Hayes, "Energy shutting hip padding system attenuates femoral impact force in a simulated fall", J. Biomech. Eng., Vol. 117, pp. 409-413, 1995a.

[14] A. J. Kroonenberg, W. C. Hayes, T. A. McMahon, "Dynamic models for sideways falls from standing height", J. Biomech. Eng., Vol. 117, pp. 309-318, 1995.

[15] L. C. Lotz, W. C. Hayes, "The use of quantitative computed tomography to estimate risk of fracture of the hip from falls", J. Bone Joint Surg.[Am], Vol. 72, pp. 689-700, 1990.

[16] C. M. Ford, T. M. Keaveny, W. C. Hayes, "The effect of impact direction on the structural capacity of the proximal femur during falls", J. Bone Miner. Res., Vol. 11, pp. 377-383, 1996.