Thermoelastic Femoral Stress Imaging for Experimental Evaluation of Hip Prosthesis Design*

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An experimental system using the thermoelastic stress analysis method and a synthetic femur was utilized to perform reliable and convenient mechanical biocompatibility evaluation of hip prosthesis design. Unlike the conventional technique, the unique advantage of the thermoelastic stress analysis method is its ability to image whole-surface stress ($\Delta(a_1 + a_2)$) distribution in specimens. The mechanical properties of synthetic femurs agreed well with those of cadaveric femurs with little variability between specimens. We applied this experimental system for stress distribution visualization of the intact femur, and the femurs implanted with an artificial joint. The surface stress distribution of the femurs sensitively reflected the prosthesis design and the contact condition between the stem and the bone. By analyzing the relationship between the stress distribution and the clinical results of the artificial joint, this technique can be used in mechanical biocompatibility evaluation and pre-clinical performance prediction of new artificial joint design.

Key Words: Biomechanics, Muscle and Skeleton, Optical Measurement, Thermoelastic Stress Analysis, Artificial Hip Joint, Synthetic Femur, Stress Distribution, Mechanical Biocompatibility

1. Introduction

Stress and strain analysis is indispensable for understanding the biomechanical function of bone and joint. It is also required for mechanical biocompatibility evaluation of artificial joints. Stress analysis of “in vitro” femur implanted with an artificial prosthesis has been performed to assess design modifications of the prosthesis. To perform the analysis, researchers have used experimental techniques such as the strain gauge method\(^{(1)-(5)}\), photoelastic\(^{(6)}\) and holographic\(^{(7)}\) techniques, as well as numerical methods such as the finite-element method (FEM)\(^{(8)-(13)}\). A crucial requirement for practical FEM analysis is the creation of the most suitable numerical model. Experimental data are also necessary for the creation and evaluation of the numerical model.

For the “in vitro” specimen experiment, the strain gauge method is the most popular technique. Although this method is useful and reliable, it is impossible to visualize whole-surface stress (strain) distribution in specimens because of number and size limitations of the gage. Visualization of stress distribution will provide much information on the biocompatibility of the prosthesis. In addition, to perform the “in vitro” cadaveric test, differences in shape and mechanical properties between cadaveric...
specimens have been another problem. Stress variation due to the modification of the prosthesis design must be distinguished from variation due to the specimen.

In order to solve these two problems for in vitro stress analysis, we employed a new experimental system that uses the thermoelastic stress analysis method and the synthetic femur. The thermoelastic stress analysis method utilizes the thermoelastic properties of materials and is a noncontact technique for whole surface stress field analysis. It enables imaging of the distribution of the sum of surface principal stresses ($\Delta (\sigma_1 + \sigma_2)$). Synthetic femurs became commercially available as substitutes for cadaveric specimens. The mechanical properties of the femurs were shown to agree well with those of cadaveric femurs with little variability between specimens\cite{14}. We applied this system for stress distribution visualization of intact femur and femur implanted with an artificial joint. The reliability and potential of the system were investigated for use as a new mechanical biocompatibility evaluation method for prosthesis design.

2. Materials and Methods

2.1 Thermoelastic stress analysis method

2.1.1 Thermoelasticity The thermoelastic effect is the change in temperature that accompanies stress change under adiabatic conditions of a body, as described in Eqs. (1) and (2). It was proposed by Thomson (Lord Kelvin)\cite{18} and the basic thermodynamics was described by Biot\cite{19}. Investigations of stress imaging using the thermoelastic effect had been performed for several years, and with the recent progress of thermal imaging technology, more precise stress visualization has become available in various industrial research fields\cite{20}. The authors have applied this technique to study cadaver bone and joint biomechanics\cite{21,22,23,24}

The equation relating temperature changes to applied stress under adiabatic conditions in a linear elastic, isotropic, homogeneous material is

$$\Delta T = -K \cdot T \cdot \Delta \sigma,$$

(1)

where $T$ is the absolute temperature of the material, $K$, the thermoelastic constant of the material, and $\sigma$, the sum of principal stresses. The thermoelastic constant, $K$, is given by

$$K = \alpha/(\rho \cdot C_p),$$

(2)

where $\alpha$ is the coefficient of linear thermal expansion, $\rho$, the density, and $C_p$, the coefficient of specific heat at constant pressure of the material\cite{25}.

2.1.2 Thermoelastic stress analysis equipment

The thermoelastic stress analysis method is based on the measurement of a small temperature difference ($\Delta T$) that occurs in a material when it is subjected to elastic cyclic loading. The stress analysis equipment consists of two units: an oscillator unit (servohydraulic testing machine, MTS 858 minibionix, MTS, U.S.A.) and a thermoelastic stress analysis unit (JTG6010, JEOL, Japan). (Fig. 1). The oscillator unit applies sinusoidal load to the material. Temperature difference ($\Delta T$) distribution in the material is detected using an infrared camera with an HgCdTe semiconductor sensor that detects $8 - 13 \mu m$ infrared wavelength. The detection of temperature change is synchronized with the load cycle with an appropriate phase difference to obtain the maximum temperature difference. Then, the temperature difference is converted to the difference of the sum of the principal stresses ($\Delta (\sigma_1 + \sigma_2)$) using Eq. (1).

2.2 Materials and experimental procedure

2.2.1 Thermoelastic property of synthetic femur

Synthetic femurs (composite femur #3103, Pacific Research Laboratories, U.S.A.) were manufactured from glass-filled epoxy. To examine the thermoelastic properties of the cortical bone of the femur, cortical bone plate of composite femur #3103 was prepared for six flat plate specimens that had a straight part 70 mm in length, 25 mm in width and 5 mm in thickness. The surface of the specimens was painted non-glossy black with aerosol lacquer paint.
(Asahi pen, Japan) in order to fix surface emissivity. Sinusoidal tensile load and compressive load were applied at 5 Hz to the specimens by the servohydraulic testing machine (MTS 858 miniplus, MTS, U.S.A.). The oscillatory amplitudes were 300 ± 200 N, 550 ± 450 N, 800 ± 700 N, 1050 ± 950 N, 1300 ± 1200 N, 1550 ± 1450 N and 1800 ± 1700 N. When each value was converted into stress difference (Δσ), the value became ± 3.2, ± 7.2, ± 11.2, ± 15.2, ± 19.2, ± 23.2 and ± 27.2 MPa, respectively. During oscillation, temperature difference (ΔT) was measured using the thermoelastic stress analysis apparatus (JTG8010, JEOL, Japan). The relationship between ΔT and Δσ was evaluated.

2.2.2 Surface stress distribution (Δ(σ1 + σ2)) change due to prosthesis implantation Synthetic femurs (composite femur #3103, Pacific Research Laboratories, U.S.A.) were used for this experiment. The distal part of the femur was fixed in a specimen holder at approximately 9 degrees of valgus in the frontal plane and neutral in sagittal plane with eight screws and acrylic bone cement. The holder was fixed in the servohydraulic testing machine (MTS 858 miniplus, MTS, U.S.A.). The apparatus applied a 5 Hz sinusoidal compressive load to the femoral head through a steel plate. As the maximum value of hip joint force during level walking was two to four times the body weight[25], an oscillatory amplitude of 1.8 kN (1.0 ± 0.9 kN) was applied. Surface stress distribution (Δ(σ1 + σ2)) images of the femur were acquired in the anterior, posterior, medial and lateral planes using the thermoelastic stress analysis apparatus (JTG 8010, JEOL, Japan). The ΔT images were obtained by conducting line accumulation eight times and frame accumulation sixteen times.

To investigate whether the method is effective for the detection of stress difference due to prosthesis design modification, four conditions of the femur were examined. The first was intact femur, the second condition was femur implanted with a rasp (for PerFix system, Kyocera, Japan), the third was femur implanted with an artificial joint (Φ22 mm ball head, Co–Cr alloy, XS SHT NK, 3M, U.S.A.) without cement fixation, and the fourth was the same artificial joint with cement fixation. The X-ray image in Fig. 2 shows the anterior view of a synthetic femur implanted with the artificial joint. The diameter of the medullary cavity was 28 mm and the stem diameter was 15 mm in the BB' direction. The diameter of the medullary cavity was 13 mm and the stem diameter was 8 mm in the AA' direction. Firm porous epoxy was filled in the medullary cavity to serve as the cancellous bone of the synthetic femur. An orthopedic surgeon performed the implantation of the femoral components. The femoral head was removed from the plane at a distance of approximately 20 mm from the lesser trochanter in this experiment. The cancellous bone was bored with a drill and rasped. Then, the stem was struck. The experiment was carried out at room temperature, about 300 K.

3. Results

3.1 Thermoelastic properties of synthetic femur

Figure 3 shows the relationship between stress difference (Δσ) and temperature difference (ΔT) of the average of six cortical bone plates. The temperature difference was measured from ΔT images by

![Fig. 2 X-ray image of a synthetic femur implanted with an artificial joint (Φ22 mm ball head, Co–Cr alloy, XS SHT NK, 3M)](image_url)

![Fig. 3 Thermoelastic property of the synthetic bone](image_url)
conducting line accumulation eight times and frame accumulation sixteen times. The negative stress value represents the compression, and the positive value, tension. Stress difference was inversely proportional to temperature difference. The proportion coefficient was about $-227 \text{ MPa/K}$.

### 3.2 Surface stress distribution ($\Delta(\sigma_1 + \sigma_2)$) change subject to prosthesis implantation

Visualization of the surface stress distribution ($\Delta(\sigma_1 + \sigma_2)$) change of the synthetic femur subject to prosthesis implantation was realized. The difference in stress reduction effect due to the difference in the design of the prosthesis was also visualized by the thermoelastic stress analysis, as shown in Fig. 4. Figure 5 indicates schematic sketch of $\Delta T$ counter map of the thermoelastic stress images.

In the medial aspect of the intact femur, compressive stress was predominantly imaged. The largest stress was found to occur around the femur neck and the proximal part. The image also indicated the presence of a tensile stress area in the distal part of the femur. In the lateral aspect, tensile stress was mainly observed in the proximal region and compressive stress was observed in the distal region of the femur. The largest tensile stress was imaged in the upper part of the femoral diaphysis.

In comparison, in the femur implanted with an artificial hip joint, marked reduction of compressive stress was imaged around the medio-proximal part of the femur. The decrease in tensile stress was also recognized in the lateral aspect. Compared with the femur implanted with the rasp and with the artificial hip joint with and without cement fixation, the surface stress pattern of the femur was altered in both medial and lateral aspects, as shown in Fig. 4. For example, the reduction of stress in the proximal part of the femur was marked under the condition of the non-cemented stem. The stress reduction effect modified by the design of the prosthesis and the interface conditions was visualized by thermoelastic stress analysis.
Natural cortical bone also exhibits a linear relationship between $\Delta \sigma$ and $\Delta T$. Vanderby and Kohles\textsuperscript{(25)} thermographically measured heat flux from loaded specimens of bovine cortical bone and correlated the results with strain gage data. The correlation coefficients demonstrated a significant linear relationship between thermal data and measured and computed stress, strain data. As the thermoelastic stress constant $K$ of natural cortical bone calculated from the physical properties is about $1.10 \times 10^{-12}$/Pa, the stress resolution becomes 3.03 MPa.

It was clarified that thermoelastic stress analysis of synthetic bone could be carried out with stress resolution equal to or higher than that of natural bone. Cristofolini et al. validated experimentally of the mechanical behavior of synthetic femurs and compared it to that of human fresh-frozen and dried-rehydrated specimens under different loading conditions\textsuperscript{(14)}. The behavior under axial loading, four-point bending test and torsional test was studied. In all of those tests, the mechanical behavior of the synthetic femurs was shown to be comparable to that of the cadaveric specimens, with no significant differences being detected between the synthetic femurs and the two groups of cadaveric femurs. Moreover, the interfemur variability for the synthetic femurs was 20 – 200 times lower than that for the cadaveric specimens. Based on these results, we concluded that the synthetic femur was a suitable specimen for thermoelastic stress analysis in order to visualize stress distribution change due to modification of prosthesis design.

In order to perform precise analysis, it is also necessary to note that the synthetic bone has no trabecula structure, and that it has a different structure of cancellous bone compared to that of natural femur.

4.2 Surface stress distribution ($\Delta (\sigma_1 + \sigma_2)$) change due to prosthesis implantation

One major problem of total hip replacement is the loss of proximal bone often found around noncemented stems in the long term\textsuperscript{(19)}. Bone loss affects the strength of the femur and increases the risk of bone fracture. Adaptive bone remodeling due to stress reduction was identified as one of the major causes of resorption\textsuperscript{(20,22)}. This was confirmed in patient studies and animal experiments, which showed that the observed effects of implant design parameters could be explained as consequences of stress shielding\textsuperscript{(29)}. If the initial stress distribution of the femur has a relationship with adaptive remodeling and long term clinical results, the detection of small changes of the stress distribution will be important for biomechanical assessment of prosthesis design. However, surface stress and strain distribution has not been sufficiently
discussed in the in vitro study by the strain gage method, due to number and size limitations of the gages. As most strain gage measurements have been carried out at 20 mm to 50 mm gage intervals\(^{(11)(12)}\), the analysis only predicts the tendency of strain distribution from a limited number of data. Unlike the conventional technique, the unique advantage of the thermoelastic stress analysis method is the high spatial resolution imaging of surface stress distribution \((\Delta T(\sigma_1 + \sigma_2))\). Surface stress distribution images sensitively reflect the prosthesis design and interface conditions between the stem and the bone, as shown in Fig. 4. The tendency of the stress pattern qualitatively coincided with the results of dried human femur\(^{(18)}\).

By analyzing and understanding the relationship between the initial surface stress distribution of the femur and the clinical results of the artificial joint, this experimental system can be used in pre-clinical performance prediction and biomechanical evaluation of new artificial joint design.

In the present study, the reduction of stress in the proximal part of the femur was marked in the case of the noncemented stem. The position of the high stress area also changed with the contact condition between the stem and the bone, as shown in Fig. 4 and Fig. 5. McNamara et al.\(^{(19)}\) reported that normal press fitting of a stem did not produce the same stress shielding effect as that obtained by gluing a press-fitted stem. They concluded that since the contact patterns were the same for both glued and nonglued stems, the greater stress shielding effect of the glued stem was probably a result of the greater initial stem-bone stability that induced a more rigid system. Rietbergen et al.\(^{(20)}\) explained that in press-fitted stems, smooth surface implants were thought to induce higher proximal bone loading and less stress shielding than bonded implants, because they were wedged into the femur every time when loaded. Although our experimental results were the same as theirs in that stress reduction was modified due to cement fixation, the tendency was opposite. Stress reduction appeared markedly in the case of the noncemented stem. In this case, the reason is considered to be that the proximal contact gap between the stem and the bone seems to exist. Since the bone around the distal part of the stem mainly supported the load due to the existence of the proximal contact gap, stress reduction in the proximal part of the femur seemed to appear. In order to confirm this hypothesis, cross sections in AA' plane and BB' plane (Fig. 2) were examined. The results showed that there was a submillimeter gap layer in entire circumference of the stem in the BB' plane. The bone cement entered there. In the AA' plane, most of the stem was direct contact with cortical bone and cancellous bone. As cancellous bone was made of firm porous epoxy and filled the medullary cavity, it contributed sufficiently to supporting the load in the distal part of the synthetic femur in this experiment. On the other hand, as the entire surface of the rasp was wedged into the bone and exhibited uniform contact, the stress distribution pattern seemed to be similar to that of the cemented fixation. By FEM analysis, Rietbergen et al.\(^{(20)}\) answered the question of how bone resorption occurred even in a press-fitted stem. Densification of distal bone bed during the initial remodeling process was found to reduce axial stem displacement and decrease the wedging effect of the stem. These, in turn, decreased the loading of the proximal bone, and then proximal bone loss occurred. In addition, it is considered from our results that initial implant conditions, such as the existence of a proximal contact gap, may accelerate the decrease of bone density in the proximal area around the press-fitted stem. In order to estimate precisely the remodeling process of the bone, it is necessary to determine the initial fixation conditions of an artificial joint. This experimental system will contribute to the investigation and confirmation of the remodeling process of bone generated by an artificial joint, by using a synthetic femur modified according to measured surface stress distribution and the remodeling theory.

5. Conclusions

A novel experimental system using the thermoelastic stress analysis method and a synthetic femur was utilized to perform reliable and convenient mechanical biocompatibility evaluation of prosthesis design. We concluded that the system was reliable and suitable for visualizing stress distribution change \((\Delta T(\sigma_1 + \sigma_2))\) due to modification of the prosthesis design. By analyzing the relationship between the stress distribution and the clinical results of the artificial joint, this technique can be used in mechanical biocompatibility evaluation and pre-clinical performance prediction of new artificial joint design.

References

(3) Indong, O.H. and Harris, W.H., Proximal Strain Distribution in the Loaded Femur, J. Bone and