OBSERVATION OF THE BLOOD FLOW IN MICROCHANNEL WITH STENOSIS BY
CONFOCAL-MICRO-PIV

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Introduction
Mass transport in human cardiovascular system takes place mainly in microcirculation, so it is important to understand blood flow in microvessels. In such small microchannels, behaviors of red blood cells (RBCs) strongly affect the flow field.

Recently many researcher developed various systems to visualize blood flow in microchannels. But it is very difficult to visualize the center plane of the blood flow when the concentration of RBCs (HCT) is over 10%. In our study, we used confocal-micro-PIV and PTV systems to overcome this problem, and investigate blood flow through stenosis.

Materials and Methods
Figure 1 illustrates the confocal-micro-PIV system used in the study. It is consists of inverted microscope (IX71; Olympus, Tokyo, Japan), confocal scanning system (CSU22; Yokogawa, Tokyo, Japan), high speed camera (Phantom v7.1; Vision Research, NJ, USA), DPSS laser (Laser Quantum, Cheshire, UK), syringe pump (KD Scientific, Holliston, MA, USA). And we used PDMS microchannel with rectangular cross section which has 100um width and 80um height. The stenosis has 50um width.

We used 3 kind of working fluid to measure the flow field: (a) pure water with 0.1% fluorescent particles (FluoSpheres carboxy1te 1.0 um orange, Sigma, UK), (b) RBCs with DEX40 (Ohtsuka-seiyaku Corporation, Japan) and 0.1% fluorescent particles, (c) 10% labeled RBCs by fluorescent dye (C-7000, Molecular Probes, USA) with DEX40. We used objective lens with 20 times magnification. The blood was taken from 23 years old man and used in room temperature. The Reynolds number based on the channel width is about 0.05 throughout this study.

Results
(1) Pure water
First, we measure the flow of pure water by the confocal-micro-PIV systems with the frame rate of 1[ms]. The results show that the flow field is similar to the Poiseuille flow before and after the stenosis, and the velocity field is almost symmetric before and after the stenosis (though the result is omitted in this paper).

(2) Red Blood Cells with fluorescent particles
The velocity field of sample (b) measured by the PIV system is shown in Fig.2. We see that there is no big difference in the flow field between sample (a) and (b).

(3) Labeled Blood Cells
The trajectories of RBCs (sample (c)) measured by the PTV system are shown in Fig.3. In the Stokes flow field of Newtonian fluid, the stream lines should be symmetric before and after the stenosis. But we found that the trajectories are no longer symmetric because of the hydrodynamic interaction between RBCs.

Discussions
From the result of the sample (a) and (b), there seems no big difference in the time averaged flow field between the pure water and 10% RBCs. This result is consistent with our prior study [1] which shows RBCs don’t affect plasma flow severely when HCT is under 10%.

In Fig.3 the trajectories are not symmetric before and after the stenosis. Therefore, we can conclude that RBC behaviour shows strong unsteadiness. This is because the RBCs interact hydrodynamically, which changes RBCs positions and orientations from when they are alone.

Conclusions
There is no big difference in the time averaged flow field between pure water and 10% RBCs suspension. But the trajectories of individual RBCs are not symmetric before and after the stenosis. Therefore we have to consider the unsteady motion of RBCs and treat Blood as a RBCs suspension diluted with plasma.

References
BLOOD FLOW DISTURBANCE BY ADMINISTRATION OF LIPOSOME IN BREAST CANCER

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The tumor tissue requires the development of tumor vessels to obtain sufficient nourishment. Recent strategies for treatment have attempted to disturb the blood flow and to achieve the definite drug delivery only to tumor tissue. We focused on the effect of large sized liposome called by Large Unilamellar Vesicles (LUV) with the size of 1μm. LUV tends to accumulate in tumor vessels and expect to disturb the blood flow. LUV is also advantageous to perform drug delivery by encapsulated drug. Increasing evidence shows that orthotopic and ectopic organ environments differentially influence tumor properties. Therefore mammary window for intravital microscopy is an important tool to study orthotopic tumor vessels, and to reveal hemodynamics and microvascular architecture. It is very useful to study chemotherapy, drug delivery, and so on. In this study, we revealed hemodynamics and microvascular architecture in orthotopic breast cancer by using mammary window and evaluate the effect of administration of LUV on blood flow. C3H mice and MCA mammary carcinoma were used for mammary windows. DiO-labeled RBC, FITC-dextran, and Rhodamine-labeled LUV were injected via tail vein. The present result shows that RBC velocity is more heterogeneous in tumor vessel compared to that of normal vessels, the tumor constitutes the fine texture network by short vessels and LUV accumulate in only tumor vessels, RBC velocity decreases after administration of LUV. In conclusion, we revealed hemodynamics and microvascular architecture in orthotopic breast cancer. LUV seems to be useful drug carrier from the result that LUV accumulates in tumor vessels and decrease RBC velocity.
Measurement of O2 and NO by the Microcoaxial Electrode

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Introduction
In addition to oxygen, nitric oxide is an important physiological substance. Particularly many previous studies have focused the function of NO for vasorelaxation and neurotransmission. Therefore we need to measure O2 and NO with high spatial and temporal resolution to reveal its physiological function in vivo. Kitamura et al. [1] showed that the oxygen microelectrode developed by Lubbers was used for the measurement of NO as an oxidation electrode. However, the limiting current was too weak for in vivo measurement. So the electrode should be more sensitive and stable for in vivo measurement. In this study, we tried to modify the fabrication of electrode for the improvement of sensitivity and stability.

Materials and Methods
Structure of Microcoaxial electrode
Fig. 1 shows the structure of microcoaxial electrode. The working pole is platinum (Pt) wire. The reference pole is sputtered silver (Ag). We modified to process the electrode tip for the improvement of sensitivity. Then a double membrane of collodion and polystyrene was placed in the tip surface for achievement of selectivity of O2 and NO. Finally, the electrode has a recess at the tip. This recess creates a concentration gradient for NO diffusion to the working-electrode surface.

In vitro measurement
We used gas-saturated saline for the performance evaluation. We used 0, 10 and 20% oxide. And we used 0 and 750 ppm NO.

O2 Measurement
The polarogram was measured in 20% O2 gas (143 mmHg) solution. We decided the appropriate voltage from the plateau area of the polarogram. We used -0.65V here.

We observed the current response to the changes of O2 concentration. Before measurement the saline was bubbled by N2 gas (0%O2 gas) for at least 30 min. And, we changed over the bubbling gas with 15 minutes interval (0% to 10% to 20% to 0%).

NO measurement
From previous study, we decided appropriate voltage of the working electrode at +0.6 V.

The saline was deoxygenated by bubbling it with Ar gas (0 ppm NO gas) for at least 30 min. And, the saline was bubbled with NO 750 ppm gas (about 1 µM NO) for 10 min and then bubbled with Ar gas for 10 min. This operation was repeated 3 times.

Results and Discussion
O2 response
Fig. 2 is response of the current to 0, 10, 20% O2. Concentration is changed at 365, 380, 395 min. Good linearity is observed. Fig 2 shows the limiting current of micro electrode for the change of O2 concentration up to 20%. We should note that the limiting current increases almost linearly with the increase of O2 concentration, which is a desirable property for measurement in vivo.

NO response
Fig. 3 shows the limiting current for the change of NO concentration up to 750 ppm. Concentration is changed at 190, 200, 210, 220, 230, 240 min. The increase of current due to the increase of NO concentration is 10 µA, which is almost ten times as large as that obtained by Kitamura et al. Although, the deformed wave form indicates poor temporal response, the sensitivity of NO appears to be feasible for the measurement of NO in vivo. Further study, we must investigate the relationship between condition of membrane and response time.

References
MEASUREMENT OF MULTI-RED BLOOD CELLS INTERACTIONS IN BLOOD FLOW 
BY CONFOCAL MICRO-PTV

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Introduction
In microcirculation the flow behavior of red blood cells (RBCs) plays a crucial role in many physiological and pathological phenomena. For instance, the interaction of RBCs in shear flow is believed to play an important role to the thrombogenesis process. Despite the relevance of this phenomenon on the blood mass transport, very little studies have been performed during the years, partly due to the absence of adequate visualization techniques able to obtain both direct and quantitative measurements on multi-RBCs motions in concentrated suspensions. Past studies on both individual and concentrated RBCs used conventional microscopes and/or ghost cells to obtain visible trace RBCs at high concentration suspension of blood cells [1, 2]. Recently, advances of confocal microscopy and consequent advantages over conventional microscopes have led to an emerging technique known as confocal micro-PTV [3, 4].

This paper presents the application of a confocal micro-PTV system to measure RBC-RBC hydrodynamic interactions in flowing blood.

Materials and Methods
Working fluids and microchannel: In this study we used dextran 40 (Dx40) containing about 20% (20Het) of human red blood cells (RBCs). All blood samples were stored hermetical at 4°C until the experiment was performed at controlled temperature of about 37°C. The microchannel was a circular borosilicate glass (100 µm in diameter).

Experimental set-up: The confocal micro-PTV system consists of an inverted microscope (IX71, Olympus) combined with a confocal scanning unit (CSU22, Yokogawa), a diode-pumped solid state (DPSS) laser (Laser Quantum Ltd) with an excitation wavelength of 532 nm and a high-speed camera (Phantom v7.1). The microchannel was placed on the stage of the microscope where the flow rate was kept constant by using a syringe pump. The confocal images were captured at a rate of 100 frames/s and then evaluated in Image J (NIH) [6] by using a manual tracking MTrackJ [7] plugin.

RBC radial displacement: The radial displacements (ΔR) of the tracked RBCs were determined by using a cumulative radial displacement, given by:

\[ ΔR = \sum [R_0 - R_i] \]  \hspace{1cm} (1)

where \( R_0 \) is the initial radial position and \( R_i \) is the cumulative radial displacement for a defined time interval.

Results and Discussion
Figure 1 shows the streamlines of two-RBC interactions around the plasma layer at Re = 0.007 (\( \gamma = 16 \, \text{s}^{-1} \)). This figure shows clearly the radial disturbance effect due to the collision with a neighboring RBC.

![Figure 1 RBC interactions (40x objective lens).](image)

Figure 2 shows the radial displacement of a RBC (RBCint) that have interacted with a neighboring RBC. Additionally, it is also shown the ΔR of a RBC (RBCnotint) with any appreciable interaction at 3% Het. These results show clearly the fluid-dynamical interactions effect on the motion of RBCs flowing in concentrated suspension of blood cells.

![Figure 2 ΔR comparison between a RBC with interactions (20% Het) and RBC with no interactions (3% Het).](image)

The present study provides both quantitative and qualitative evidence on RBC-RBC hydrodynamic interaction in flowing blood at Het up to 20%. The measurements were possible due to the unique ability of the confocal systems to obtain thin in-focus planes. Hence, this confocal micro-PTV system can provide the paths of two or more RBCs interacting in the same focal plane. Such information is extremely important to elucidate the blood transport mechanisms and associated diseases such as thrombosis and atherosclerosis.

References
QUANTITATIVE MEASUREMENT OF DISTRIBUTION OF RED BLOOD CELLS IN MICRO FLOW USING MICRO PTV

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Introduction
Red blood cell (RBC), a major component of blood, causes unique phenomena in microcirculation such as axial migration because of several distinct characteristics of RBC id est biconcave shape and high deformability [1]. These phenomena have been discussed both experimentally in vivo [2] and numerically [4]. However, the influence of each characteristic of RBC and the distribution of RBC have been considered in the experiment of the deformability of RBC on hemorheology, the distribution and velocity profiles of both real RBC and hardened RBC (HBC) in microchannel were measured by micro particle tracking velocimetry (PTV).

Measurement System
The micro PTV system consisted in a fluorescent microscope (DMIRE2, Leica Microsystems), high-speed camera (Phantom v7.1, Vision Research Inc.), and a high NA objective lens (PL APO CS 63x/1.40-0.60 HCX, Leica Microsystems). This system enables minimally invasive measurement.

Experimental Condition
The experimental condition for RBC and HBC are shown in Table 1. To neglect the interaction between the cells, haematocrit was set to 1%. Physiological saline was used as working fluid. To measure the velocity of fluid, φ0.5 μm polystyrene fluorescent particles were added in the working fluid at 0.2% volume fraction. The fluid was constantly injected by syringe pump at 30 μl/h to microchannel which was made of polydimethyl-siloxane. The shape of the channel was straight whose cross section was rectangular with 50 μm(W) x 62 μm(H). The Reynolds number was 0.14 which is same with that of arteriole. RBC or HBC and tracer particles were separately observed at 1, 20, 50, 100, 200, and 300 W from the entrance of the channel. The measurement area was 90.6 μm x 22.7 μm and the camera frame rate was 1000fps.

Table 1 The properties of RBC and HBC

<table>
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<th>Animal</th>
<th>Size</th>
<th>Fluorescent label</th>
<th>Haematocrit</th>
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<tr>
<td>RBC</td>
<td>Sheep</td>
<td>5 μm</td>
<td>DIO</td>
</tr>
<tr>
<td>HBC</td>
<td>Goat</td>
<td>4.5 μm</td>
<td>φ100 nm polystyrene particle</td>
</tr>
</tbody>
</table>

Analysis method
For detection of RBC or HBC, binarization method with dynamic threshold [5] was applied. RBC is detected correctly with this method although fluorescent intensity of dyed RBC is low and RBC imaging shape changes continuously. Detection of defocused RBC which is out of measurement depth was avoided by considering standard deviation of luminance value. For tracer particles, particle mask correlation method was applied.

Results and Discussion
The probability density functions (PDF) of cells distribution are shown in figure 1. The distribution of HBC is almost flat at the center of the channel. On the other hand, RBC concentrates at the center. Since the physical differences between RBC and HBC are considered the deformability of RBC and tank tread motion, these characteristics are thought to drive the axial migration of the cell. Nevertheless, there are no significant difference in the velocity profiles between RBC and HBC in this experimental condition.

Conclusions
We measured the distribution and velocity profiles of RBC and HBC in the micro flow using micro PTV in order to clarify the effect of RBC deformability. We clarified that RBC migrates the center of the channel more than HBC, and no significant difference between RBC and HBC were observed in velocity profiles.

References
TRANSPORT MODEL OF PHYSIOLOGICAL SUBSTANCES IN CORTICAL BONE THROUGH POROUS CANAL NETWORK

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Introduction

Bone changes its microstructure according to the mechanical environment. It is suggested that mechanical stimuli are sensed by osteocytes, and osteocytes regulate the activity of osteoblasts and osteoclasts for bone microstructure change according to the degree of mechanical stimuli [1]. Although it is considered that physiological substances (PSs) transport in bone matrix is important for maintaining the activity of these cells, the detailed mechanism is not well understood yet. In cortical bone, PSs are mainly supplied via the vascular channel network constituted by Haversian and Volkmann’s canals, and by canaliculi network constituted by bone lacunae and canaliculi. The purpose of this study is to model transport phenomena of PSs in cortical bone by considering these two networks. The model validity was demonstrated by calculating the PS distribution around a single vascular canal.

Transport model of physiological substance

Transport phenomena of physiological substances in cortical bone was modeled by considering the transport through Haversian and Volkmann’s canals and the transport through bone matrix continuum having homogenized canaliculi network. In Haversian and Volkmann’s canals, transportation of PSs was formulated as the equation of one-dimensional advection and diffusion along individual canal taking the substances exchange with the bone canaliculi into account. In the bone matrix, it was formulated as the equation of three-dimensional diffusion with diffusivity equivalent to that in bone canaliculi network taking the substances exchange with the canal network into account. Thus, time derivatives of the concentration of PSs in the canal $C^0$ and that in bone matrix $C$ are written by

$$\begin{align*}
\frac{\partial C^0(s,t)}{\partial t} &= -R_g \frac{\partial C^0(s,t)}{\partial s} + D_i \frac{\partial^2 C^0(s,t)}{\partial s^2} \\
&- \frac{2}{R_i} \frac{2}{D_i} C^0(s,t) - C^i(s,t), \\
\frac{\partial C(x,t)}{\partial t} &= \frac{1}{V_F(x)} \int_{R_i} D_i \frac{\partial^2 C(x,t)}{\partial x^2} \\
&- \frac{2 \pi R_i D_i}{Q} \left[ C^0(s,t) - C^i(s,t) - Q \right]
\end{align*}$$

(1)

(2)

where $i$ and $j$ indicate junction number of the canal network. In these equations, $D_i$ and $R_i$ are the mean flow rate and the mean radius of the vascular vessel connecting the junctions $i$ and $j$, $D_i'$ and $D_j'$ are diffusion coefficients in plasma and vascular wall, $i$ is the thickness of vascular wall, $V_F$ is the ratio of bone volume and volume of lacunae and canaliculi, $L$ is the Euclidean norm of direction cosine normalized by its maximum component of canal, $Q$ is the consumption rate of physiological substance per unit volume, and $D_i'$ is the diffusion coefficient in bone matrix derived from the fiber matrix theory used in references[2,3].

Fundamental analysis of transport model

For verification of validity of this transport model, the distribution of PS concentration was analyzed around a single vascular canal in a cubic bone matrix (Fig. 1(a)), where are the concentration gradients vertical to the boundary surfaces are set to be zero. The PS distributions at equilibrium in cross section ABCD are represented in Fig. 1(b), (c) and (d), where $Q_i$ is the consumption rate of $O_i$ for basal metabolism. The $C_i$ concentration decrease as consumption rate increase(Fig. 1(b), (c)), and the substance deflection area (i.e., $C_i=0$) appeared as consumption rate particularly increased(Fig. 1(d)). Thus, it may considered that the transport analysis of PS based on the proposed transport model make it possible to discuss mechanism of substance supply to osteocyte.

Fig. 1 Substance concentration at steady state. (a) Single canal model. (b) Cross-section ABCD. $Q=Q_i$ (c) Cross-section ABCD. $Q=10Q_i$ (d) Cross-section ABCD. $Q=10^2Q_i$

References


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TARTRATE-RESISTANT ACID PHOSPHATASE-POSITIVE CELL DIFFERENTIATION INFLUENCED BY OVER-PHYSIOLOGICAL STRETCHED OSTEOCYTES


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Introduction
The sites of microdamage must be sensed and promptly remodeled to avoid leading to clinical bone fractures. Osteocytes inside bone matrix should play an important role in detecting the microdamage. The aim of this study was, therefore, to develop a mechanical loading apparatus that could apply both physiological and over-physiological strain to the gel-embedded osteocyte, mimicking the induction of microdamage in bone matrix. To examine usefulness of the apparatus, the gel-embedded MLO-Y4 cells were subjected to mechanical stretching with wide strain range, and then the potential of MLO-Y4 cells was evaluated by adding culture supernatant to undifferentiated bone marrow cells.

Materials and Methods
An apparatus for stimulating the cells was newly designed to stretch an elastic culture well. The elastic culture well (Figure 1) was molded with silicone elastomer. The culture well consisted of 10 separated culture spaces, and could apply 5 different strain magnitude to the cells at the same time.

For mimicking in vivo environment, osteocyte-like cell line MLO-Y4 was embedded in collagen gel and cultured three-dimensionally (3-D) in vitro. After allowing the base layer to solidify by a short incubation, 200μl of the cell-suspended collagen-Matrigel mixture was poured into each culture well (2 x 10⁵ cells/ml of gel) [1].

The relationship between the displacement of the movable rod and generated strain inside the gel was preliminary evaluated by measuring deformation of gel-embedded cells. Mounting the elastic culture well on the stage of a microscope, location of the Hoechst-stained cells was monitored with a CCD camera. When the well was stretched from 0 to 3mm by 0.25mm step in displacement, fluorescent microscopic images of the stretched cells were obtained in each region of interest (Figure 1).

To examine the influence of mechanical stretching to cell viability, distribution of dead cells was assessed in the gel-embedded MLO-Y4 culture after 24-h mechanical loading. The culture wells were mounted on the loading apparatus, and then the cells were subjected to 24-hour intermittent stretching at the frequency of 2Hz. The cells were respectively suffered from stretching of 933, 1780, 2240, 2640 and 3000με in strain pattern I, and 4670, 8890, 11200, 13200 and 15000με in strain pattern II. After the application of mechanical stretching, the cells were stained with 2mM ethidium homodimer-1 (EthD-1) for identifying dead cells.

The gel-embedded MLO-Y4 culture forming 3-D cellular network was prepared as described above. The cells were subjected to the stretching regime of pattern I and II, respectively. After the application of mechanical stretching, supernatants of culture media were collected and mixed with fresh medium at the concentration of 20%. These conditioned media (CM) were immediately used for the following bone marrow cell culture. Bone marrow cells were obtained from ICR mice. Non-adherent bone marrow cells were collected and then applied into a 24-multi well plate with the CM at the density of 10⁵ cells/well. After a week of incubation, cellular tartrate-resistant acid phosphatase (TRACP) activity was evaluated.

Results and Discussion
Linear relationship was observed between the rod displacement and generated cell-to-cell strain. According to this strain measurement, tensile strain up to 48,800με or 4.8% could be applied to the cells when the rod was displaced from 0 to 3mm. Strain magnitude in well number 5 was 3.21 times as large as that in well number 1.

Viability assay indicates that significant dead cells were observed by the application of stretching over 5000με (Figure 2C and D). Beyond this threshold, the application of mechanical stretching increased the number of dead cells in a magnitude-dependent manner, and significant amount of the cells was injured by the stretching over 8890με (Figure 3).

MLO-Y4 cell supernatant suffered from the strain beyond the threshold at 8890με showed a significant increase in TRACP activity (Figure 4). These experimental findings indicate that local death of osteocytes provides an important mechanism to target bone resorption to microdamage site.

Conclusions
A mechanical loading apparatus that could apply both physiological and over-physiological strain to the gel-embedded osteocyte was developed. Over-physiological stretching mechanically induced cell death in the gel. The supernatant of the damaged osteocytes had a potential to induce TRACP-positive cells in bone marrow culture.

References

Figure 1 Elastic culture well having 10 separated culture spaces.

Figure 2 Dead cell distribution after mechanical stretching. Non-stretched (A), 3000με (B), 5000με (C), 15000με (D). Bar: 100μm.

Figure 3 Number of dead cells induced by mechanical stretching.

Figure 4 TRACP activity normalized by non-stretched group.
Evaluation of Mechanical Property and Micro-structure of Neck of Femur of Osteoporotic Rat using Scanning Acoustic Microscope


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Introduction
A bone fracture in the neck of femur inflicts a serious problem on the elderly person because of a high possibility of being bedridden. An external reason of the fracture is mainly fall, while internal reason of it is osteoporosis due to ageing. The elderly who live in cold and snowy area are particularly liable to slip and fall on the snow covered road during the winter. Furthermore, the frigid stress from the cold weather is pointed out to be concerned with progress and appearance of osteoporosis. In this study, the characteristics of micro-structures and mechanical properties of normal bone and osteoporotic bone of rats were investigated by using a scanning acoustic microscope. A second moment of inertia and area ratio of trabecula structure were evaluated by using C-mode measurement and an image processing.

Materials and Methods
Normal femurs were obtained from five normal rats (Wister Kyoto Rat), and osteoporotic femurs were obtained from five osteoporotic rats (Stroke Prone Spontaneously Hypertensive Rat). These femurs were cut along the long axis as shown in Figure 1. The cut femurs were embedded in epoxy resin. Then the surface of specimens were polished with buffing machine. The characteristics of micro-structures and mechanical properties were investigated by using a scanning acoustic microscope (SAM) with a frequency of 200MHz. Young’s modulus of them were analyzed along the measurement positions set on the neck of femur (cf. Figure 1) via distributions of leaky surface wave velocity and acoustic impedance of microscopic area of bone tissue measured by SAM. A second moment of inertia and area ratio of trabeceula structure were evaluated by using C-mode measurement and an image processing. A rectangular coordinate system, taken the X axis in the direction of long axis, was set on the specimen surface. Then U and V axes were defined as parallel X and V axes on the center of gravity of trabecula. Second moments of inertia of trabeceula, I_U and I_V, were analyzed around U and V axes, respectively.

Results and Discussion
Young’s modulus distribution of osteoporotic bone is shown in Figure 2. Average value of Young’s modulus of osteoporotic bone was 8.2±4.2GPa, while that of normal bone was 10.6±6.0GPa. The Young’s modulus of osteoporotic bone was slightly lower and more homoscedastic than that of normal bone. The area ratio of osteoporotic trabecula was similar with that of normal bone. Second moment of inertia of trabecula is shown in Figure 3. The second moment of inertia I_U of osteoporotic trabecula was smaller than that of normal trabecula, while I_V of osteoporotic trabecula was larger than that of normal trabecula. It seems that osteoporotic trabecula is structured slenderly along long axis of femur.

Conclusions
Mechanical properties and micro-structure of neck of femur of osteoporotic rat were investigated by using SAM. The results are as follows. Mechanical properties of osteoporotic cortical bone and trabecula were slightly lower than those of normal bone. Area ratio of trabecula of osteoporotic bone was lower than that of normal bone. The trabecula along the direction of long axis of osteoporotic bone was more slender than that of normal bone.

Figure 1 Cross section of neck of femur of rat

Figure 2 Young’s modulus distribution along the line across the neck of femur of osteoporotic bone

Figure 3 Second moment of inertia of trabecula
THE MECHANICAL PROPERTIES OF TRABECULAR BONE IN KNEE JOINT

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Introduction
In this study, we performed the compressive strength test of trabecular bone in knee joint for measuring the elastic modulus and compressive strength. In the case of osteoarthritis, some patients have only medial condylar osteoarthritis[1]. So we performed the mechanical test for comparison the difference of strength each condyle.

Materials and Methods
We used 26 knees from 11 cadavers (7 females, 4 males). The averaged age was 80 years for females, 64 years for males. The medio-lateral angle of knee was measured by X-ray images. The value of knee angle was 177.7±1.8°. All knees had varus.

Femurs and tibias were harvested by surgical method. We never used thermal or chemical treatments because these actions make a change the mechanical properties. Distal femur and proximal tibia were sliced by linear precision saw (ISOMAT 5000, Buehler, IL) and cored by diamond core drill for making cylindrical specimens. We obtained six specimens from distal femoral part and two specimens from proximal tibial part. (figure 1)

Figure 1 Coring position and specimen configuration.

The size of specimens was 12mm diameters for distal femur, 15mm diameters for proximal tibia. The heights of specimens were 1.5-2.0 times of diameters[2]. The specimens were compressive tested after had been made as soon as possible. We used universal test machine (5567, Instron, MA) for performing the compressive test. The specimens for test was submerged 38°C normal saline.

Results and Discussion

<table>
<thead>
<tr>
<th>Bone</th>
<th>Sex</th>
<th>Location</th>
<th>Ultimate Strength[MPa]</th>
<th>P Value*</th>
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<td></td>
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<th>P Value*</th>
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<td>F</td>
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<td>2.91±1.26</td>
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<td>Middle</td>
<td>2.01±0.79</td>
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<td>Posterior</td>
<td>4.24±1.38</td>
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<td>Average</td>
<td>3.05±1.47</td>
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<td>M</td>
<td>Anterior</td>
<td>7.40±1.95</td>
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<td>Middle</td>
<td>4.86±1.01</td>
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<td>Average</td>
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<td>5.96±2.83</td>
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<td>F</td>
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<td>1.06±0.41</td>
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<td>M</td>
<td>Middle</td>
<td>1.54±0.78</td>
<td>3.66±0.97</td>
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* F: Female, M: Male

Conclusions
We obtained the bellow results about the mechanical properties of trabecular bone in knee joint for Korean.

Mechanical properties of distal femur
The elastic modulus was 361.1±159.41MPa for males and 150.31±70.64MPa for females. The compressive strength was 6.78±2.90MPa for males and 2.87±1.41MPa for females. Males have 2.4 times more strong than females about elastic modulus and compressive strength.

Mechanical properties of proximal tibia
The elastic modulus was 108.82±52.88MPa for males and 73.47±55.06MPa for females. The compressive strength was 2.60±1.38MPa for males and 1.76±1.16MPa for females. Males have 1.5 times more strong than females about elastic modulus and compressive strength.

Comparison of mechanical properties medial with lateral condyle
Lateral condyle was strong than medial condyle at anterior and posterior region (LA, MA, LP, MP) and medial condyle was strong than lateral condyle at middle region (LM, MM) about the strength at the trabecular bone in distal femur. But there were no difference by statistical analysis at significance level 0.05 except for middle region in females. The other regions (anterior, posterior, male, female) have no difference of strength. But, medial region was always more strong than lateral region in proximal tibia.

References
STRUCTURAL EFFICACY OF LOCAL ADMINISTRATION OF A BISPHOSPHONATE ON NECROTIC CANCELLOUS BONE

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Introduction

Ischemic osteonecrosis of the femoral head (IOFH) is one of the diseases that affect cancellous bone. Although its etiology is unclear, osteonecrosis occurs by deprivation of the blood supply to the femoral head. This bone ischemia leads to structural failure of cancellous bone, which results in the collapse of the femoral head. This structural failure is the result of resorption of the necrotic bone, and around viable bone during revascularization, before new bone has formed and consolidated sufficiently to become load-bearing [1].

If osteclastic bone resorption is suppressed after osteonecrosis, collapse can be prevented or delayed. Recently, systemically administered bisphosphonates have been used to treat osteonecrosis and showed a short-term clinical effect in delaying femoral head collapse [2, 3]. However, since IOFH is a localized condition, there is a need to explore the therapeutic potential of local administration of bisphosphonate to prevent femoral head deformity. In the present study, we investigated the structural efficacy of local bisphosphonate administration on necrotic cancellous bone.

Materials and Methods

We evaluated the morphological changes of necrotic cancellous bone that was immersed in a bisphosphate solution during bone regeneration in the bone conduction chamber (BCC). BCC has been used for studies of bone regeneration [4]. It consists of two half-titanium screws and a hexagonal closed screw cap with a cylindrical interior space (Fig. 1). The interior space has a diameter of 2 mm and length of 7 mm. One end of the chamber is screwed into the bone. BCC has two small bone ingrowth openings that are located at the bone end (Fig. 1).

Structurally intact cancellous bone grafts were obtained from 14 female 6-week-old Sprague Dawley rats. A cylindrical 2 × 3 mm cancellous bone rod was resected in the axial direction from the tibial cancellous region under the growth plate. The grafts were kept sterile and frozen at -70°C. These allografts were used as a model for necrotic, autologous cancellous bone. Half of the grafts (n = 7) were immersed in a bisphosphate solution. Two 5-mg tablets of alendronate (Fosamax, Banyu Pharmaceutical, Japan) were dissolved in 10 mL of water. After thawing, the grafts were placed in the bisphosphate solution (1 mg/mL) for 10 minutes and then rinsed three times in saline, to remove unbound alendronate [5]. Thereafter, grafts were placed in BCC and implanted unilaterally in SD rats (10 weeks old, 389 g). Non-immersed grafts (n = 7) were treated as controls. After 3 weeks, the rats were killed and the contents of the chambers were scanned using a micro-CT.

A three-dimensional dataset was reconstructed from the multiple slice image data. A 2-mm sphere located at the bone ingrowth side was set as a volume of interest (VOI). The regenerating cancellous bone grafts were evaluated using conventional morphometric parameters of trabecular microstructure. All data obtained from the morphometry were analyzed statistically using the Mann-Whitney U-test. All animal experiments were conducted according to the “Guidelines for Animal Experimentation” of the Niigata University Graduate School of Medical and Dental Sciences.

Results and Discussion

Figure 2 shows the contents of the BCC 3 weeks after implantation. Soft tissue invaded the grafts from the bone ingrowth holes (Fig. 2A), which shows that vascularization has occurred in the bone chamber. In the controls, the necrotic cancellous bone grafts were partly resorbed and a large marrow cavity was observed (Fig. 2B). In contrast, the bisphosphonate-treated grafts appeared to be intact and new bone formation was observed (Fig. 2C).

The trabecular microstructure parameters are listed in Table 1. Significant differences were observed in the bone volume fraction (BV/TV), trabecular number (Nb), trabecular separation (Tb.Sp), and bone surface (BS/BV). These results show that bisphosphonate-treated necrotic cancellous bone is not resorbed with increased new bone formation. The bisphosphate inhibited necrotic bone resorption by adhering closely to necrotic bone surfaces and inactivating osteoclasts that would resorb the bone. Therefore, the trabecular microstructure was maintained, and even strengthened, by new bone formation on the remaining necrotic bone surface.

In conclusion, our findings indicate that local bisphosphonate administration prevents necrotic bone resorption during revascularization after osteonecrosis and maintains the trabecular microstructure. Therefore, this treatment may prevent femoral head collapse in patients with osteonecrosis of the femoral head.

References


Figure 1 Bone Conduction Chamber (BCC).

![Bone Conduction Chamber (BCC)](image)

Table 1 Structural parameters

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Control</th>
<th>Bisphosphate</th>
</tr>
</thead>
<tbody>
<tr>
<td>BV/TV, %</td>
<td>30.3±2.6</td>
<td>49.6±14.8</td>
</tr>
<tr>
<td>Th.Th, mm</td>
<td>0.04±0.009</td>
<td>0.06±0.02</td>
</tr>
<tr>
<td>Th.N, 1/mm</td>
<td>6.15±0.89</td>
<td>7.90±0.31</td>
</tr>
<tr>
<td>Th.Sp, mm</td>
<td>0.11±0.02</td>
<td>0.06±0.01</td>
</tr>
<tr>
<td>BS/BV, mm²</td>
<td>41.38±6.72</td>
<td>34.31±9.85</td>
</tr>
<tr>
<td>DA</td>
<td>1.20±0.06</td>
<td>1.20±0.10</td>
</tr>
</tbody>
</table>

Mean ± S.D. values * p < 0.05 vs. control.
Inelastic Constitutive Modeling of Cortical Bone Taking Account of Anisotropic Damage and Viscoelasticity

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Introduction

In the previous paper [1], one of the authors formulated a constitutive model for cortical bone by taking account of anisotropic elastic coefficients with strain rate dependency, strength anisotropy, strength asymmetry of tension and compression, and isotropic damage. In the present paper, we improve the model by taking into account viscoelasticity and anisotropic damage.

Formulation of a Constitutive Model

The proposed constitutive model is based on a simple one-dimensional rheological model. The model consists of a generalized Maxwell element and a viscoplastic element. These elements are connected in series. Thus, we represent the total strain $\varepsilon$ as the sum of the viscoelastic strain $\varepsilon^v$ and viscoplastic strain $\varepsilon^p$ in the form

$$\varepsilon = \varepsilon^v + \varepsilon^p$$

(1)

The Helmholtz free energy $\Psi$ is defined as the sum of the viscoelastic part $\Psi^v$ and the viscoplastic part $\Psi^p$

$$\Psi(\varepsilon^v, D, A, r, \gamma_1, \cdots, \gamma_N) = \Psi^v(\varepsilon^v, D, A, \gamma_1, \cdots, \gamma_N) + \Psi^p(D, r),$$

(2)

where $D$ is a second-order symmetric damage tensor, $A$ is the structural tensor expressing transverse isotropy with respect to the axis of long bone, $r$ is an isotropic hardening variable in viscoelasticity, and $\gamma_i$ ($i = 1, \cdots, N$) is internal state variables for viscoelasticity. From Clausius-Duhem inequality, a viscoelastic equation and the relations between internal variables and their associated thermodynamic conjugate forces $(\sigma, \dot{\sigma}, \dot{Q}_1, \cdots, \dot{Q}_N)$ can be obtained.

Then we apply the hypothesis of total energy equivalence in the damage mechanics to the free energy $\Psi$ to take into account the damage effects. According to the hypothesis, the damage effects are taken into account by replacing the arguments of the free energy by the corresponding effective variables $(\mathbf{\varepsilon}^v, \mathbf{\dot{r}}, \mathbf{\dot{\gamma}}_1, \cdots, \mathbf{\dot{\gamma}}_N)$ with $D = 0$.

The dissipation potential $F$ is defined as the sum of the flow potential $F^f$ and the recovery potential $F^r$.

$$F(\sigma, A, \dot{\sigma}, \dot{A}) = F_f(\sigma, A, \dot{\sigma}) + F_r(\dot{A})$$

(3)

In this study, we adopt the Tsai-Wu criterion [2] as a yield criterion of cortical bone. The evolutions of viscoplastic strain $\varepsilon^p$ and isotropic hardening variable $r$ are derived from normality rules.

By assuming that the principal axes of the damage tensor $D$ coincide with those of the structural tensor $A$ during damage process, and by using the representation theorem [3], we formulated the evolution equation of the form

$$D = \dot{D}_I + \dot{D}_r (I - A)$$

(4)

where $\dot{D}_I$ and $\dot{D}_r$ denote an isotropic damage evolution and a rapid damage evolution in the transversely isotropic plane, respectively. These coefficients are formulated as follows:

$$\dot{D}_I = a_I \left( \frac{f}{K} \right)^{\frac{1}{n}} \left( \frac{tr(\sigma^v)}{\sigma^v} \right)^{\frac{1}{m}} \exp(C_I trD)$$

(5)

$$\dot{D}_r = a_r \left( \frac{f}{K} \right)^{\frac{1}{n}} \left( \frac{tr(I - A)\sigma^v}{tr\sigma^v} \right)^{\frac{1}{m}} \times \left( 1 + b_r \frac{tr(\sigma^v)^{\frac{1}{m}}}{tr\sigma^v} \right) \exp(C_r tr(I - A)D)$$

(6)

where $a_I, b_r, C_I, a_r, b_r, C_r, n, m$ are material constants, $f$ is the equipotential flow surface, $K$ is the coefficient of plastic resistance, and $\sigma^v$ is the yield stress. These equations can represent transverse isotropy and differences in tension and compression in deformation and strength.

Applicability of the Proposed Model

To confirm the validity of the proposed constitutive model, we simulated the experimental data in the literature [4] by the proposed model. The experiments are the uniaxial tensile and compressive tests along the axis of the transverse isotropy and in the transversely isotropic plane of human femur cortical bone. Figure 1 shows that the simulation can predict the experimental data with a good accuracy.

![Fig.1 Comparisons of simulation result with the corresponding experimental data [4]](image)

Conclusion

The proposed model is requisite to clarify some problem in impact biomechanics, such as the prediction of bone fracture pattern. For this purpose, the model will be incorporated into a FEM Code in near future.

Reference

A COMPARATIVE STUDY OF ANTI-RESORPTIVE AND ANTI-CANCER TREATMENTS IN THE GEOMETRICAL AND BIOMECHANICAL PROPERTIES OF RAT FEMURS WITH TUMOR-INDUCED OSTEOLYSIS

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While much has been learned about the mechanisms of metastatic spread of cancer to bone, there has been little headway in establishing guidelines for monitoring the response of metastases to treatment. The objective of this study was to investigate the efficacy of two different treatments (Anti-Resorptive:ibandronate and Anti-Cancer:Paclitaxel) of tumor induced osteolysis in terms of geometrical and biomechanical parameters.

A rat femur model for tumor-induced osteolysis using W256 cancer cells was adapted from a previous study. Of the 30 rats implanted with cancer cells, 12 were untreated (CANCER), 9 received Ibandronate (IBAN), and 9 received Paclitaxel (PAC). Another 12 underwent a sham operation (CONTROL). Micro-computer tomography (µT) scans of the femurs extracted post-mortem were used to quantify standard geometrical parameters such as bone volume (BV) and cross-sectional area (CSA). A 3-point bending test was used to assess the femurs' mechanical stiffness. Finally, serum levels of deoxypyridinoline (Dpd) were measured to obtain the degree of bone resorption, which in turn gave an indication of cancer activity.

CSA data (taken at 25% of the total femur length from the distal end) from the CANCER group had the largest % difference between the left and right femurs (corresponding with a 13.4% increase in Dpd levels). IBAN rats showed the least % difference in CSA (38.1% drop in Dpd levels), followed by PAC rats. This pattern is repeated in the % differences in BV. The anti-cancer effect of PAC can be inferred from the lower Dpd levels of the PAC group (8.86% increase) as compared to the CANCER group. Both IBAN (2.85%) and PAC (-3.86%) maintained the stiffness properties of the operated femurs, with PAC being marginally more effective.

The initial results suggest the viability of this rat model for bone metastasis, and that IBAN was most effective in maintaining skeletal geometry. Interestingly, PAC seemed to be more capable of preserving the biomechanical properties of the femur, despite deterioration in its geometrical properties. A subsequent full scale study, with additional bone mineral content and histomorphological analysis, is in progress to confirm these results.
NUMERICAL ANALYSIS OF AIRFLOW IN A HEALTHY HUMAN AIRWAYS

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** Department of Otorhinalaryngology, Dalian Medical University, Dalian. Liaoning. 116027 China

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The quantificational measure of velocity and pressure distribution in human airway can hardly be carried out directly due to the complex and fragile anatomical structure. While mechanical modeling is a feasible method to quantitatively explore the airflow field characteristics in airways during respiration. It is beneficial to investigate the rapacities, toxicology and risk assessment protocols for human healthiness. In order to describe the airflow field characteristics, airflow through an anatomically representative three-dimensional model of the human airway has been numerically simulated using computational fluid dynamic (CFD) technique and the finite element (FE) method. The airway model which was built based on the CT images of a healthy volunteer and Weibel's symmetric model, consists of two connected segments, the airways from the nostril to the trachea (Generation G0) and an upper tracheobronchial three model of G0-G3. It is comparatively true to reflect the actual anatomical configuration. The airway walls were assumed to be passive and rigid, and the non-slip boundary conditions were used on the interior walls of the airways. Considering oscillatory turbulent incompressible three-dimensional flow in the airways, airflow fields were solved for a cycle of quiet breathing with a tidal volume of 0.6L and a cycle of 3s. Computational fluid dynamics (CFD) results are quantitatively determined. Significant variations of the airflow apportionment occurred in the posterior nasal valve region, laryngeal vestibule and bifurcation of trachea. And the regions are prone to induce some abnormal phenomena owing to the enhancement of the airway compliance. Airflow field of velocity and pressure distribution across the airway generally agreed with experimental results from the literatures.
NUMERICAL ANALYSIS OF THE MECHANICAL PROPERTIES IN NORMAL AND DISEASED LUNG USING A SINGLE ALVEOLAR DUCT MODEL


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Introduction
To understand lung ventilation in normal and diseased states, mechanical behaviour of alveoli must be described. Recently, there were few attempts to make numerical model to tackle the mechanical behaviour of lung parenchymal tissue. The proposed models, however, require massive computing power to predict the behaviour or recover limited behaviours. Therefore, in this study, a simple, but powerful alveolar duct model comprised of a flexible spherical cell in pressurized vessel and the equations of motion are proposed in consideration of the viscoelastic forces of the collagen and elastin fibers and the surface tensile forces of the air-liquid interfaces.

Models and Methods

Figure 1 Compartment alveolar duct model.

It is assumed that the air flow in branches is laminar and obeys Poisuelle’s law and the relationship between transpulmonary pressure and the total force in alveolar wall is described by the law of Laplace. In order to obtain the pressure-volume behaviour of the communicating alveolar duct model, breathing is driven by a sinusoidal change of the total volume with time. For each step, volume and pressure of either alveolus are determined such that the external force due to the given total volume change is balanced by viscoelastic forces and air-liquid interface surface tensile forces.

The surface tension of the air-liquid interface on the surface of the alveolar walls is described using the model developed by Otis et al. [2]. The nonlinear time and history dependent viscoelastic behaviour of the parenchymal biological tissue is described using the quasi-linear viscoelastic theory by Fung [1]. Also, the properties of collagen and elastin fiber bundles are modelled by the method of Denmy et al. [4].

Results and Discussion

To understand the mechanisms of alveolar behaviour, it is investigated the dynamic properties of the model which are the tissue resistance ($R_t$) and dynamic elastance ($E_{dyn}$) (Fig. 2). The $R_t$ decreased hyperbolically with frequency and decreases linearly as tidal volume increases. The $E_{dyn}$ increases with increasing frequency and falls linearly with increasing tidal volume, as do the tissue resistance, in agreement with previous experimental observations [3].

The distinguishing features of lung diseases such as pulmonary emphysema are changes in airspace dimension and tissue properties. Figure 3 shows the P-V curves of the model where one compartment alveolus has half of the dynamic stress response of the normal. In some respects, this result is pertains to emphysematous lung and is in agreement with previous observations. Destruction and remodelling of alveolar tissue in emphysematous lungs result in a loss of elastic recoil, which is reflected by a shift to the left of the entire P-V curve.

Conclusions

In this study, alveolar duct model is proposed that describes the mechanical behaviour of parenchymal tissue very powerfully. It is presumed that the model is to be an index which explores various theoretical mechanisms such as asynchronous ventilation increasing bolus dispersion observed in several diseased lungs.

References

Figure 2 Dependence of alveolar duct model dynamic property on frequency.

Figure 3 Quasi-static P-V curves of alveolar duct model. Solid line, both normal alveoli; Dashed and dotted lines represent normal and affected alveoli by tissue destruction, respectively.
Monitoring physical and mental workloads by a wearable computer


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Introduction

Japan and Korea, like other developed countries, has a rapidly aging society. Consequently, the demand for home care is growing rapidly, and the most important issue in medicine is supporting elderly patients both physically and mentally. In this reason, we have proposed the Hyper Hospital Network as a technological environment to reorganize medicine for patients. The Hyper Hospital environment that we are developing covers patients’ homes, hospitals, local health care networks, etc., and involves various interconnected computers with server functions or peer-to-peer network.

The purpose of this study is that developing a wearable system to monitor an elderly people by ambulant measurements of some vital signs non-invasively. We examined some physiological measurements such as the heart rate variation and the pulse wave velocity and got power spectrum of R-R interval. With these data we could analyze patients' autonomic nervous condition which can be influenced by physical and mental workloads.

Methods

ECG and pulse wave velocity were measured in-house developed equipments using a PSoC(Programmable System on Chip) and an accelerometer(APA-300) (Figure 1). As an experiment, eight healthy male volunteers were subjected to the measurements under some of physical and mental workloads. Experimental process was to put the subjects in an order of (1) rest for 10 min., (2) manual calculation for 10 min., (3) rest for 10 min., (4) physical exercise - riding a bicycle for 5 min., (5) rest for 10 min., (6) mental and physical repose – listening classical music for 10 min., and (7) rest for 10 min.. ECG and pulse wave signals were transformed to digital data with a sampling rate of 120Hz/channel with 14bit resolution.

Results and Discussion

Figure 2 shows the pulse-waves those were measured from sensors and accelerometer. Because femoral artery’s pulse-wave (upside) and ECG (downside) were acquired simultaneously, the velocity of the pulse wave propagation from heart to femoral region can be calculated with interval between two peaks. The R-R intervals obtained from ECG were analyzed by using power spectrum analysis method. Power spectrum was split into the LF (low frequency: 0.04-0.15Hz) and the HF (high frequency: 0.15–0.4Hz) components [1-2]. In this analysis, heart-rate, PWV, power of LF and HF components and ratio of LF and HF component which is widely used as an index of sympathetic activity were calculated to analyze the variance of each process.

Table 1 Variation in each process (where avr. Indicated average variation rate of 8 subjects)

<table>
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<th>Variation</th>
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<tr>
<td>Mental workload</td>
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<tr>
<td>(1) → (2)</td>
<td>PWV increase (avr. 3%)</td>
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<tr>
<td></td>
<td>LF increase (avr. 47%)</td>
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<td></td>
<td>LF/HF increase (avr. 27%)</td>
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<tr>
<td>Physical workload</td>
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<tr>
<td>(3) → (4)</td>
<td>Heart-rate increase (avr. 40%)</td>
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<tr>
<td></td>
<td>PWV increase (avr. 14%)</td>
</tr>
<tr>
<td></td>
<td>LF decrease (avr. -80%)</td>
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<tr>
<td></td>
<td>HF decrease (avr. -88%)</td>
</tr>
<tr>
<td>Mental &amp; Physical repose</td>
<td></td>
</tr>
<tr>
<td>(5) → (6)</td>
<td>Heart-rate decrease (avr. -15%)</td>
</tr>
<tr>
<td></td>
<td>LF/HF decrease (avr. -5%)</td>
</tr>
<tr>
<td>Repose to normal state</td>
<td></td>
</tr>
<tr>
<td>(6) → (7)</td>
<td>Heart-rate increase (avr. 3%)</td>
</tr>
<tr>
<td></td>
<td>LF/HF increase (avr. 31%)</td>
</tr>
</tbody>
</table>

Conclusions

In this study, we developed a module of a wearable computer which can measure important bio-information including heart-rate, PWV and power of LF and HF components. It was possible to detect mental loads and physical loads of people by overall analysis of collected information. We expect that this system will be adapted to Hyper Hospital Network and be helpful for elderly home care.

References

ANALYSIS OF CALCIUM CONTENT IN HUMAN ARTERY

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Introduction
Arterial calcification is a major cause of arteriosclerosis, and has been considered to be a diagnostic index of the progression of arteriosclerosis. However, the mechanism of calcification in artery is not clear. To clarify the mechanisms of vascular calcification, detailed distribution of calcium in artery is necessary. Radiological image analysis of artery is a facile method for measuring calcification, but the quantitiveness of this method is not sufficiently validated[1]. Additionally, in conventional studies of calcium distribution in artery, characteristics of shape of arteries such as bifurcation and bend was focused, and simple factors of artery shape were related to distribution of calcium in artery. As real shape of artery contains multiple factor, the effect of artery shape on calcium distribution has been insufficiently investigated.

In this study, we analyzed calcium content in human arteries dissected from Japanese and Thai by using radiological images and verified the radiological image method by comparing with atomic emission spectrometry (AES), which provides the high-precise analysis of calcium content. Thereby we measured calcification in human internal artery and middle cerebral artery with multiple bend, and analyzed the distribution of calcification.

Materials and Methods
Sampling of Artery: Cadavers were treated by injection of a mixture of 36 % ethanol, 13% glycerine, and 6% formalin through the femoral artery. After the ordinary dissection by medical students was finished, carotid, coronary, internal carotid and middle cerebral arteries were resected. The arteries were washed thoroughly with distilled water to remove uninterested tissue and blood clot and used for the determination of calcium contents. For calcium determination by AES, the artery was divided into cylindrical segments.

Determination of calcium content: The samples were dried at 80 degrees centigrade for 16 h. After the addition of 1 ml nitric acid, they were heated at 100 degrees centigrade for 2 h in a dry block bath (EB-303, Iuchi). After the addition of 0.5 ml perchloric acid, they were heated at 100 degrees centigrade for a further 2 h. The samples were adjusted to a volume of 10 ml by adding ultrapure water and filtered through filter paper with a pore size of 4 µm. The resulting filtrates were analyzed with an inductively coupled plasma-atomic emission spectrometer (ICPS-1000, Shimadzu). The amounts of elements were expressed on a dry-weight basis.

Radiological image protocol: A X-ray CT scanner (ELE SCAN, NITTETSU ELEX CO.) and a radiographic X-ray equipment (smx1000, Shimadzu) were used for acquisition of CT and transfer images respectively.

CT images of Coronary arteries and carotid arteries were scanned with slice thickness of 1 mm. The scan parameters were tube current 100 µA, tube voltage 60 kV, and 256-gray scale images were obtained. X-ray transfer images of internal carotid and middle cerebral arteries were scanned with 5.0 of magnification ratio.

Calcification index was defined as mean density in the artery area in the radiological images.

Results and Discussion
Validation of quantification in radiological method: Figure 1 shows the correlation between calcium content measured by AES and the calcification index defined by the mean density of X-ray CT images. The calcification index was positively correlated with the calcium content, and the results demonstrate that CT image analysis is effective for the calcification analysis.

Site-dependence of calcification: Figure 2 (a) shows an X-ray CT image of carotid artery where calcium locally accumulated. AES has insufficient spatial resolution for detecting the localized calcification due to sample preparation. In contrast, X-ray image analysis has advantage to measure detailed distribution of calcium.

X-ray transfer image of internal carotid artery and middle cerebral artery (Figure 2 (b)) showed that middle cerebral artery was more calcified than internal carotid artery. Because both these arteries have similar bending shape, this deference in calcification suggests importance of multiple shape factors.

Conclusions
Calcium distribution was complicated radially and axially. To understand the mechanism of vascular calcification, X-ray image analysis is available.

Figure 1 Correlation between calcium content measured by atomic emission spectrometry and mean density of X-ray CT images.

Figure 2 Example of CT image (a) and transfer image (b) of internal and middle cerebral artery.

Reference
DESIGN OF MULTILAYERED COCULTURE SYSTEM COMPOSED OF SMALL HEPATOCYTES, LIVER STELLATE CELLS AND SINUSOIDAL ENDOTHELIAL CELLS

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Introduction
The liver’s elaborate architecture consists of heterotypic cell interactions for the phenotypic stability of the parenchymal cells as well as for the proper liver function. Particularly liver tissues exhibit layered cellular architectures, and it is desirable to reconstruct the multi-layered cell culture systems for the achievement of tissue-engineered liver.

Recently we achieved to form the bile canaliculi by cultured small hepatocytes (SHs) and hepatic stellate cells (HSCs) on the collagen coated plate. To reconstruct more elaborate architecture of liver tissue, we demonstrated the design of multilayered coculture system composed of SHs, SECs and HSCs to mimic the characteristic 3D relationship of the these cells within the microenvironment such as space of Disse and elucidated validity of the multilayered coculture system.

Materials and Methods

Cell Isolation: The liver cells (SHs, HSCs, and SECs) were isolated from male SD rat using the two-step liver perfusion methods.

Cell Culture: SHs were seeded on the lower part of the semipermeable polyethylene terephthalate (PET) porous membrane (25 μm thick, 1.0×10⁵ pores / cm²) with 8.0 μm-sized pores in the culture insert. SHs were cultured in DMEM supplemented with 10% FBS, dexamethasone, insulin, nicotinamide, ascorbic acid, EGF, transferrin, and antibiotics. for 14-20 days. After cultivation of SHs for 2-3 weeks, SECs were seeded on the upper part of the membrane. After the seeding of SECs, 10ng/ml vascular endothelial growth factor (VEGF) was added to the medium.

Imaging of coculture system: SECs, HSCs, and SHs in the coculture system were localized using immunofluorescence staining. Digital images of the fluorescence distribution of the cells were obtained using confocal laser-scanning microscope (CLSM). ELISA for SHs albumin secretion: The albumin secretion by SHs were measured by enzyme-linked immunosorbent assay.

Results and Discussion

Migration of HSCs in Coculture System: Before plating SECs, we observed the localization of HSCs by immunofluorescence staining. The number of HSCs detected on the upper part of the membrane was increased with the time course of the culture days. These results suggest that HSCs cultured on the lower part of membrane migrated to the upper part through the micropores.

Involvement of SHs for maintenance of SECs in coculture system: The cell number of SECs was maintained for coculture system as compared with the monoculture models. The cell number decreased with time in all conditions. Especially, in monoculture model without VEGF, cell number decreased rapidly from day3 to day5. Finally, SECs could not be observed day6. Similarly in monoculture model with VEGF, cell number decreased from day2 to day3 rapidly. On the other hand, in coculture model, the cell number was maintained at high ratio until day5. The high surviving ratio of SECs in the coculture system indicates the importance of heterotypic interactions with SHs and/or HSCs.

Correlation between SECs and HSCs in Coculture System: Double-immunofluorescent staining of SECs and HSCs in the coculture model revealed that piled-up aggregates composed of 6-10 SECs located on the desmin-positive HSCs. These results suggest that SEC-HSC interaction is involved in the formation of SEC aggregates in the coculture model.

Conclusions
In this study, we demonstrated the coculture system consisting of SHs, HSCs, and SECs, the major components of hepatic tissues. Although further investigation will be needed, this coculture system will be useful to investigate heterotypic cell interactions, which is important for tissue organization in vitro.

Figure 1 Migration of HSCs in Coculture System.

Figure 2 Correlation between SECs and HSCs in Coculture System.

Figure 3 Maintenance of Albumin Secretion in Coculture System.
Effect of shear stress on three-dimensional microvessel network formation of endothelial cells

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Introduction

Angiogenesis appears for the growth of tumor tissue and the disturbance of angiogenesis may control the growth of tumor tissue. On the other hand, the construction of angiogenesis is desirable for the treatment of the ischemic heart disease. Furthermore the vascular network formation is indispensable for the functional maintenance of reconstructed cellular structure in the field of tissue engineering.

Because the function of endothelial cells are enhanced by the mechanical stimulus of shear stress, microvessel network formation is also promoted by shear stress stimulus [1]. However, the shear dependency of microvessel formation remains unknown. So we focused on the microvessel formation dependent on the shear stress stimulus. In this study, we observed the microvessel formation with laminar shear stress from 0.1 to 2.0 Pa and discuss the effect of shear stress on the morphology of microvessel formation, particularly three-dimensional structure.

Materials and Methods

In vitro three-dimensional model made by Montesano et al [2]. Collagen gels were prepared as follows: 8 vol of type I collagen solution (3.0 mg/ml; Cellmatrix Type I-A, Nitta Gelatin) were mixed with 1 vol of 5×DMEM and 1 vol of 0.1 N NaOH (Wako Pure Chemicals Industries) on ice. The mixture was then poured into a glass-base dish and allowed to gel at 37°C for 30 min. Shear stress experiments were performed using bovine pulmonary microvessel endothelial cells (BPMEC, Cell Systems). The endothelial cells were seeded onto 1.53-mm-thick collagen gels at 4 × 10^5 cells per 35-mm culture dish to make a microvessel formation model. And these cells were cultured in Dulbecco’s modified Eagle’s medium (DMEM, Gibco) supplemented with 10% fetal bovine serum (JRH Biosciences) and 1% antibiotic-antimycotic (Gibco). The cells reached confluence 72 h after seeding, and reconstructed the immature three-dimensional network formation.

Subsequently, collagen gels with three-dimensional networks were placed into a parallel-plate flow chamber, and the endothelial cells grown on these collagen gels were subjected to well-defined laminar fluid shear stress by the flow of DMEM adding 30 ng/ml basic fibroblast growth factor (bFGF). Flow of DMEM was provided by a sterile continuous-flow loop. The perfusate was circulated by a roller pump (model MP-3N, Eyela) in the flow circuit. Shear stress is set the following measure by a roller pump; 0.1 Pa, 1.0 Pa, 1.5 Pa, and 2.0 Pa.

Phase-contrast images of the networks formed in collagen gels were recorded at 5-min intervals for 48 h with a microscope equipped with a time-lapse system.

Results and Discussion

BPMECs were exposed to laminar shear stress from 0.1 to 2.0 Pa. The endothelial cells at the confluent state were treated with bFGF (30 ng/ml) for 48 hours. Fig. 1 show network formation of phase-contrast image under shear loaded condition for 48 hours. In Fig.1 the edge of microvessel image was intensified to have a clear process of network formation. The direction of flow is left to right. At the shear stress of 0.1 Pa (Fig. 1A), the network formation appears to be enhanced particularly after 24 hours elapsed. At the shear stress of 1.0 Pa (Fig. 1B), the network tends to grow in the flow direction. At the shear stress of 1.5 Pa (Fig. 1C), the lumen of microvessel seems to increase and the network formation seems to become slow at the shear stress of 2.0 Pa (Fig. 1D). Thus the network formation tends to be enhanced at the lower shear stress, but tends to be slow down at higher shear stress. Furthermore we should note that the microvessel was elongated significantly at lower shear stress such as 0.1 Pa (Fig. 1A), whereas the microvessel at higher shear stress appears to be short. Thus we should take into account the enhancement effect of microvessel formation at lower shear stress for the application of tissue engineering.

Fig. 1. In vitro model of 3-D network formation.

The microvessel network formation on the endothelial cell monolayer exposed by the shear stress stimulus of 0.1 Pa (A), 1.0 Pa (B), 1.5 Pa (C), and 2.0 Pa (D). Bar, 100 micron mm.

References

Deformation Analysis of Periodontal Tissue using Digital Image Correlation Analysis

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Introduction
Studies on deformation and movement of teeth have widely been performed to understand the mechanism of occlusion in the field of dental science. Previous studies were however focused on measurements of local displacement and strain at single points [1] and deformation analysis of dried teeth samples [2] and therefore, enough information for occlusion mechanism has not been obtained yet.

In the present study, global deformation analysis of a periodontal tissue prepared from a flesh skull bone of a swine was performed by means of digital image correlation analysis. Furthermore, finite element analysis (FEA) of a three-dimensional model of the tissue was conducted to characterize the mechanical behavior of the tissue.

Materials and Methods
Mandibular bone was removed from a flesh skull bone of a swine, and thin samples of 2 mm thick consisting of molar, periodontal ligament and alveolar bone were cut out of the mandibular bone using a low-speed cutter. A lateral surface of the specimen was then dried so that a color spray can be painted, and a rondam pattern was created on the surface.

Compressive tests were then conducted to simulate a fundamental occlusion process. Compressive force was applied to the top of the tooth crown, and deformation process was captured by a CCD camera. This mechanical testing was controlled by displacement, and the maximum displacement of 200μm was applied to the specimen. The deformed specimen was photographed every 20μm displacement. The displacement fields in x- and y-direction were then evaluated from each of the images by using the digital image correlation technique.

A three dimensional finite element model was also developed based on a image of a undeformed specimen. The FEA model developed is shown in Fig.1. 4-nodes tetrahedral elements were used, and the numbers of elements and nodes were 52136 and 12871, respectively. The mechanical properties of the three different tissues were assumed to be linear elastic. The material constants are shown in Table 1. The bottom ends of the FEA model were fixed totally, and the forced displacement of maximum 200 μm was applied to the top of the molar. Then, x- and y-displacements were analytically evaluated.

Results and Discussion
Distribution patterns of y-displacement obtained from the digital image correlation analysis and FEA are shown in Figure 2. These displacement fields correspond to the final displacement of 200 μm. It is important to note that for both results, the molar was mainly displaced and the displacement of the alveolar bone was very small. This clearly indicates that the deformation of the periodontal ligament connecting the molar and the alveolar bone was very large.

Conclusions
Digital image correlation analysis and FEA of deformation field of a periodontal tissue of a swine under compressive load were performed. Both results of y-displacement field exhibited good agreement, suggesting simple compressive loading condition is well predicted by FEA with linear elastic properties of tissues.

References

Table 1. Material properties

<table>
<thead>
<tr>
<th>Material</th>
<th>Young's modulus[MPa]</th>
<th>Poisson's ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Molar</td>
<td>15000</td>
<td>0.3</td>
</tr>
<tr>
<td>Periodontal ligament</td>
<td>1</td>
<td>0.49</td>
</tr>
<tr>
<td>Alveolar bone</td>
<td>17200</td>
<td>0.3</td>
</tr>
</tbody>
</table>

Figure 1 Boundary condition.

Figure 2 y-displacement field of periodontal tissue.
BIOMECHANICAL EVALUATION OF THE POST DIMENSION ON ENDODONTICALLY TREATED FIRST PREMOLAR

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Introduction

Teeth loss due to severe caries or trauma was often observed in clinical dentistry. Endodontically treatment, including canal debridement, shaping and root filling, would be performed as the cavity was restored. However, these endodontically treated teeth might have less stiffness because of the loss of sound teeth. Basically, dentists would adopt post-and-core system to rehabilitate root canal and provide retention for crowns. The post dimension was the major biomechanical concern of clinical application. Generally, preservation of more sound teeth would be benefit to provide clinical crown with decreasing the stress. Several studies were showed controversial conclusions [1][2]. This might be resulted from the variability of experiment setup. Finite element analysis (FEA) was convinced as a powerful approach for biomechanical research. As the result, this study was to investigate the effect of post dimension on endodontically treated premolar using finite element analysis.

Materials and Methods

A fresh premolar with no caries or other complications was selected for construction of the finite element (FE) model. The contour of the premolar was obtained by reverse engineering. Laser scanning was performed to get the 3D coordinates of the outer surface of the premolar. A surface model was constructed by these coordinates in the CAD environment (Solidworks 2006, solidworks co., USA). The interior structures, such as post and core system, dentine and pulp cavity, were established by relative anatomical positions. The alveolar bone structures, including cortex and cancellous bone, were assumed as a rectangular block with appropriate width and height (Figure 1(a)). The entire solid model was transferred into FE package (ANSYS 10.0, ANSYS Inc., USA). The 3D mesh model of the entire premolar was completed by mapping mesh techniques (Figure 1(b)). Material properties were assumed as isotropic materials in this study and listed in Table 1. All exterior nodes in the distal and mesial surfaces of the alveolar bone were fixed in all directions in this study. The vertical force of 200N was applied respectively at the central fossa of the premolar while oblique (45 degrees with buccal inclination) force of 200N was applied on the buccal cusp. Different post lengths (6mm, 7mm, 8mm and 9mm) and diameters (1.2mm, 1.5mm and 1.8mm) were investigated.

Table 1 Material properties used in this study

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
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<tbody>
<tr>
<td>Enamel</td>
<td>84000</td>
<td>0.31</td>
</tr>
<tr>
<td>Dentin</td>
<td>18600</td>
<td>0.31</td>
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<tr>
<td>Periodontal ligament</td>
<td>68.9</td>
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<tr>
<td>Gutta-percha</td>
<td>0.69</td>
<td>0.45</td>
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<tr>
<td>Cortical bone</td>
<td>13700</td>
<td>0.30</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>1370</td>
<td>0.30</td>
</tr>
<tr>
<td>Zirconium</td>
<td>112000</td>
<td>0.33</td>
</tr>
</tbody>
</table>

Results and Discussion

With the decrease of the post diameter, the stress was increased. When oblique force was applied, the high von Mises stress on the remaining dentine was showed at the buccal side of the enamel-dentine junction (Figure 2). The similar trend between the post length and von Mises stress was also happened under this loading.

Conclusions

From the results of this study, it showed that the post length has to longer than 8mm to lower the stress on the remaining tooth. Oblique force might be harmful to the remaining tooth structures because it resulted in higher stress which might be related to the tooth fracture.

References

EFFECT OF THE NUCLEAR ELASTICITY ON THE MECHANICAL BEHAVIOR OF AN IMAGE-BASED SINGLE ENDOTHELIAL CELL MODEL: FINITE ELEMENT MODELING AND ANALYSIS

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Introduction

A quantitative evaluation of cellular strains and stresses is important in understanding the mechanical effect on the transmission of mechanical stimuli and cytoskeletal remodeling \(^{(1)}\). Caille et al. measured the deformation of the nucleus in endothelial cells under the conditions of 25% uniaxial substrate stretch \(^{(1)}\). They quantified the elasticity of the cytoplasm and nucleus through compression testing, and carried out a finite element analysis using an axisymmetric model of neo-Hookean material \(^{(2)}\). Yamada et al. numerically analyzed the strain state in stretched cells and compared it with actin cytoskeletal structures remodeled by cyclic stretching \(^{(3)-(5)}\). These previous studies did not consider an actual shape or nuclear elasticity.

In this study, we carried out a finite element analysis to estimate the strain distributions of an image-based single endothelial cell while taking into account the nuclear elasticity.

Methods

Sliced images were obtained from a single endothelial cell on stretched and unstretched substrates by confocal laser scanning microscopy (CLSM) with an equal vertical interval. A three-dimensional (3D) geometry of the unstretched cell was reconstructed, using Igor Pro ver. 5J (Wave Metrics) and Photoshop ver. 7 (Adobe Systems) software to process the images.

Using Rhinoceros ver. 3 (Robert McNeel & Associates), we constructed a 3D geometry of the whole cell from the fluorescence images and added an assumed nucleus. The cellular height was 5.5 μm. The substrate was 40.0 μm in the x-direction by 32.9 μm in the y-direction. The nucleus had a maximal diameter of 11.7 μm in the horizontal plane. These geometrical models in IGES format were entered into Abaqus/CAE (ABAQUS), and the numerical simulations were carried out using Abaqus/CAE (ABAQUS) software. The cytoplasm, nucleus, and substrate were modeled as neo-Hookean material. The strain energy density function was expressed as

\[ W = C(I_1 - 3) \]  

where \( I_1 \) is the first invariant of the right Cauchy-Green tensor. Material constant \( C \) was set to 850 Pa and 129 Pa for the cytoplasm and nucleus, respectively, based on the measurement by Caille et al. \(^{(2)}\), and 129 kPa for the substrate. A finite element model of an image-based cell, taking into account the nuclear elasticity, was chosen a 10-node modified tetrahedron hybrid with linear pressure and hourglass control (C3D10MH). The finite element model had 13,463 elements. We applied uniform deformation to the substrate in the form of 15.1% tensile strain in the x-direction and 3.0% compressive strain in the y-direction.

Results and Discussion

Figure 2 shows the distribution of the nominal strain in the stretching direction at a cross section in the x-z plane of the cell. Comparing the strain distribution of the model with the cytoplasm elasticity in the nucleus region (not shown), we observed that the strain was smaller except at the neighbors of the nucleus around the height level of the maximum diameter of the nucleus. For example, the strain in the top region of the cell was about 2% for the nucleus elasticity case while it was about 7% for the cytoplasm elasticity case. The strain in the region just below the nucleus was about 11% for the nucleus elasticity case and 15% for the cytoplasm elasticity case. In the experiment by Caille et al., the nucleus deformation ranged from 3% to 7% when the substrate was stretched by 15% \(^{(1)}\). Our nucleus deformation result of 7% was within their measurement range, although they did not measure the geometry of the cell in their study.

In large strain regions in the cytoplasm, the remodeling of actin stress fibers may be enhanced by applying a larger mechanical stimulus to the actin cytoskeletal network.

Conclusions

In this study, we carried out finite element analysis of an image-based cell, taking into account the nuclear elasticity. The nucleus with a large elastic modulus causes smaller strains in the stretching direction in most regions of the cytoplasm, and larger strains in some small regions. Such a non-uniform strain field may affect the strain transmission or cytoskeletal remodeling.

References


Figure 1 Finite element model based on CLSM images

Figure 2 Distribution of the nominal strain in the stretching direction at a cross section in the x-z plane of the cell
ESTIMATION OF MUSCLE FORCES IN BIOMECHANICAL MODEL OF HAND WITH CONSIDERATION OF INTERACTION AMONG FINGERS

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Introduction

The biomechanical model of fingers has been studied since 1970s. The model of fingers and muscles was developed and the muscle forces as well as the joint forces according to the posture of fingers were predicted [1-3]. However, the number of the fingers involved in the model was small and in addition, the interaction among fingers was not considered though the interaction such as the enslaving effect and the force deficit effect during multi-finger force-production tasks can make the control of fingers more stable has been investigated.

In this paper, a three-dimensional biomechanical model of four fingers including three joints and the muscles was developed and the muscle forces were estimated by the optimization technique and the mathematical relationship among the fingers.

Materials and Methods

In index, middle, ring, and little fingers, there are three bones (distal phalanx, middle phalanx, proximal phalanx) and three joints (distal interphalangeal joint, DIP; proximal interphalangeal joint,PIP; meta carphophalangeal joint, MCP). The muscles in fingers can be divided into two groups which are mainly involved in extension and in flexion. Terminal extensor (TE), radial band (RB), ulnar band (UB), extensor slip (ES), and long extensor (LE) are muscles concerning with the extension and flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), intrinsic muscle of hand (INT), lumbrical (LU), radial interosseous (RI), ulnar interosseous (UI) are muscles concerning with the flexion.

Three force equilibrium equations of muscle forces, joint forces, and the external forces and three moment equilibrium equations of moments derived from those forces around a joint were formulated. Since some muscles are divided into several muscles and some muscles are combined into one muscle, the equations representing the relationship among muscles, especially given in the ratio of PCSA were formulated. However, the number of unknowns (the values of muscle forces and the joint forces) still exceeds that of equations, so that an optimization scheme was used in order to solve this redundant problem. Sequential quadratic method was utilized to the optimization scheme by Matlab(The MathWorks, USA).

From a mathematical formulation between neural commands and finger forces which represents the enslaving effect and the force deficit effect during the multi-finger force-production tasks [4], the external forces acting on the end of four fingers respectively could be calculated while four fingers push a plate at the same time. In order to compare our result with the previous literature, the flexion angles of three joints were fixed to 30, 40, and 10 degrees for DIP, PIP, and MCP, respectively.

Results and Discussion

When fully pressing a plate under the flexed posture, the muscles, FDP in DIP, and RI and UI in MCP concerning with the flexion were mainly activated regardless of fingers (Table 1). The mainly activated muscles as well as the magnitudes of the muscle forces in this paper agreed with those in the previous two-dimensional models of one finger. However, it was found that the antagonistic muscles were also activated rather than the previous models, which is more realistic phenomenon. Moreover, since the developed model has considered the interaction among fingers, this model can be more powerful while developing a robot hand that can totally control the multiple fingers like human. Also, the present study could be useful to rehabilitate the patients with problems in nervous system and to design an interface between the control command and the movement of the robot.

Table 1 Predicted muscle forces when fully pressing a plate under the flexed posture (Unit: N)

<table>
<thead>
<tr>
<th>Finger</th>
<th>Index</th>
<th>Middle</th>
<th>Ring</th>
<th>Little</th>
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<tbody>
<tr>
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<td>4.9</td>
</tr>
<tr>
<td></td>
<td>FDP</td>
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<td>8.2</td>
<td>6.3</td>
</tr>
<tr>
<td>PIP</td>
<td>RB</td>
<td>0.0</td>
<td>0.0</td>
<td>1.4</td>
</tr>
<tr>
<td></td>
<td>UB</td>
<td>2.0</td>
<td>2.3</td>
<td>1.0</td>
</tr>
<tr>
<td></td>
<td>FDS</td>
<td>2.0</td>
<td>2.2</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>ES</td>
<td>3.3</td>
<td>2.5</td>
<td>2.5</td>
</tr>
<tr>
<td>MCP</td>
<td>RI</td>
<td>4.0</td>
<td>0.0</td>
<td>4.3</td>
</tr>
<tr>
<td></td>
<td>LU</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>UI</td>
<td>6.0</td>
<td>7.5</td>
<td>5.2</td>
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<td></td>
<td>LE</td>
<td>0.0</td>
<td>0.0</td>
<td>0.0</td>
</tr>
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</table>

References


Acknowledgement

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A FINITE ELEMENT MODEL OF THE M. LEVATOR ANI BASED ON MRI
FOR THE PELVIC FLOOR MUSCLES

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Introduction:
The human pelvic floor is a very complex muscular structure. The m. levator ani with its facial covering constitutes the pelvic diaphragm and is largely responsible for supporting both pelvic and organs in the abdominal cavity. The pelvic floor dysfunction(PFD) is a kind of injury disease of the pelvic floor, which effects seriously the health of medial and aged female. The study aims to create a complete geometry of the m.levator ani using MRI and a finite element model and use it to study its complex biomechanical behavior based on the finite element theory.

Methods:
A 28 year old young nulliparous female volunteer with no known pathologies of the pelvic floor, who has the erotic experience, underwent supine MRI. By employing the digital image process technique the 3D configuration for one of the most important parts of the pelvic floor, the m.levator ani, was reconstructed. The m.levator ani was assumed to be hyperelastic, isotropic and homogeneous, and the mesh adopted the 4-nodes tetrahedra elements and the material property parameters have been defined by implementing a UMAT routine. It was assumed that the nodes connected to the pelvic bone are fixed completely and all other nodes are free.

Results:
Both 3D configuration and finite element model of the m.levator ani were established. The numerical simulations of the deformation of the m.levator ani have been performed when applying a pressure and contracting the muscle. and the preliminary simulation results of the complex biomechanical behavior of the m.levator ani were obtained.

Conclusion:
The established 3D configuration based on MRI shows the actual anatomy of the m.levator ani. It is a reliable way to build 3D finite element model with MRI data. The maximum principal stress in the element occurs along the fiber directions and the maximum stress occurs at the sling attachment points of the m.levator ani, where most post-partum lesions occur. Medical evidence seems to corroborate this. We hope this study be able to give insight into the function of pelvic floor muscles and provide suggestion for preventing and treating PFD.
DEVELOPMENT OF AN ARM MODEL WITH A BIONIC ELBOW JOINT

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Introduction
The human elbow joint has two degree of freedom. Especially, wide range for the rotation of the forearm (Pronation-Supination) is attained by the sophisticated complexity of the human elbow joint. The elucidation of its mechanism is useful for medical evaluation and application to welfare device.

The purpose of the study is to develop the physical arm model that imitates the musculo-skeletal system of human elbow joint. In this paper, we made a bone model by rapid prototyping, and evaluated the mobility of the arm model by the moment arm.

Materials and Methods
Human elbow joint has two movements. Flexion-Extension is basic movement, and Rotation (Pronation-Supination) is the movement of the forearm about its longitudinal axis.

The forearm consists of the ulna and the radius. In the position of supination, the ulna and the radius are almost parallel. In the position of pronation, the radius has intersected forward of the ulna.

In this study, firstly we made a model (Figure 1 left side) as follows;
1. The ulna and the radius are simple columns.
2. The ulna and the radius are parallel.

Then referring to the concept of ‘the shape of forearm bones has “spiral” structure’ [1], we curved the radius model (Figure 2 right side).

In the curved part, we allocated the wire as the pronator teres referring to the location and function of human muscle [2]. The pronator teres is the muscle for pronation movement.

In this model, the main bodies of the bones were made by epoxy resin as a photopolymer, and the joint parts were made by the acrylic material. The polyethylene fiber was used for the wire. The size of the model is about 3/4 of human bone.

Measuring wire excursion in flexion-extension and supination-pronation, the moment arm was calculated.

Results and Discussion
In this model, we defined as flexion angle is 0 degree when the elbow joint is extend completely, and pronation angle is 0 degree when the forearm is supinated completely (the ulna and the radius are parallel).

The range of motion of pronation in column model was 60 degree. In curved model, the range of motion became 90 degree.

The moment arm of pronation by pronator teres in each flexion angels is shown in Figure 2. The moment arm of the curved model increases as pronation angles increases, and become larger than the column model.

In our previous study [3], the mobility of 6-wire model for arm model composed of parallel columns with elbow joint was evaluated. In next step, we will curve not only the radius model but also the ulna model.

Conclusions
We made a physical model of the forearm bone by rapid prototyping, and evaluated the mobility of this arm model by the moment arm.

References
Kinematical Modeling of a Manipulative Workspace of Human Hands

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Introduction

Clinically, to assess the range of motion of the hand usually depends on the conventional goniometric measurement. Although it is simple and convenient to manipulate, its accuracy and objectivity is still questionable. In our previous studies, the methods to evaluate the maximal workspace of the fingers and thumb have been developed [1,2]. However, an actual functional working range is not equal to a maximal workspace from a clinical viewpoint. Thus, the purpose of this study is to define and compute a functional workspace of a normal hand based on the motion analysis and numerical calculation techniques. The proportion of the functional workspace of each finger relative to its maximal workspace was obtained.

Materials and Methods

10 normal subjects with no history of surgery or injury of the hand were recruited in this study. Informed consent was obtained after describing the experimental procedure. Each subject was asked to position the elbow joint in 90 degree and wrist joint in neutral position using a special designed device. Then, the subject was asked to perform three maximal movements of the thumb and fingers [1,2]. Afterwards, the functional workspace which was defined as the combined area formed by the maximal workspace of the thumb and each finger was obtained.

The maximal motion trajectories of the thumb-tip and fingertips were captured using an Eagle Digital Motion analysis system (Motion Analysis Corporation, Santa Rosa, CA, USA). The overlap regions between the maximal thumb and each fingertip workspace were determined as following procedures: (1) to pre-filter the raw motion data and interpolate the markers inside the closed envelope of each fingertip trajectory by the B-spline method; (2) to generate triangles to fill out this closed region and remove the redundant ones (Fig. 1); (3) to calculate the surface area of the fingertip workspace by summing all areas of the triangles; (5) to determine fitting surfaces of the upper and side boundaries of the thumb that are crossed by the fingers (Fig. 2); (6) to find the triangles whose three vertexes are located between the two triangles to fill out this closed region and remove the redundant ones (Fig. 1); (3) to calculate the surface area of the fingertip motion.

Results and Discussion

The graphic display of the intersection region of one finger and the thumb was shown in the Figure 4. The areas of the maximal workspace and the intersection region both were computed. Afterwards, the results of the proportion of the functional workspace of each finger relative to its maximal working ranges were listed in the Table 1. The ratio of the functional workspace (4483 mm²) and maximal workspace (25782 mm²) of a whole normal hand was 17.38%.

Conclusions

In the future, we plan to use this concept to develop a “normal functional workspace” database. It is intended to be the basis for a standard in the use of handicapped authentication, insurance claim and criteria of the return to work. Through this standardized approach, many of the artificial causes which may affect an objective decision will be avoided. Otherwise, this technique will also provide a concept of using the quantitative method for clinical occupational therapists, hand therapists or clinicians and inspire them to create more quantitative assessments to replace qualitative ones in clinical uses.

References


Table 1 – The proportion of the functional workspace of each finger relative to its maximal working ranges

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<thead>
<tr>
<th>Subject</th>
<th>Index</th>
<th>Middle</th>
<th>Ring</th>
<th>Little</th>
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<tr>
<td></td>
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<td>FW (mm²)</td>
<td>%</td>
<td>MW (mm²)</td>
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* MW: maximal workspace  *FW: functional workspace
POSTURAL CONTROL MECHANISM DURING WHOLE-BODY REACHING IN STROKE PATIENTS

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INTRODUCTION
Center of pressure (COP) trajectory is one of the most common methods used to characterize the balance control of both normal subjects and patients with neurological disorders, such as stroke. Voluntary movements, such as targeted reaching, are practical essence for daily function and have been proved to be more effective in challenging and training the balance ability than simulated balance perturbations such as moving platforms [1,2]. Seated reaching for targets in different directions at the level of shoulder height has been shown to affect COP trajectory significantly. Whole body reaching (WBR) requires the subjects to pick up a target on the floor and the target distances were found to affect the movement of COP and activation patterns of postural muscles [3]. The purposes of this study were to compare the performance of WBR with different balance ability and the effects of target distances (10% body height vs 30% body height measured from the midpoint of both big toes) and directions (middle, M; left, L) on COP trajectory as measure by COP path excursion (WTP), COP maximum displacement in frontal (MML) and sagittal (MAP) direction.

METHODS
Fifteen normal adults and 23 stroke subjects who fulfilled the inclusion criteria participated in this study. They were instructed to pick up a light weighted bean bag on the floor in two directions (in the middle and to the left or parietic side) at two distances (away from the big toe for 10% and 30% of body height) while standing erectly on R SSCAN pressure mat. A total of 12 trials (3 x 2 x 2) were required. The functional reach distance (FR) was also measured as an indicator of balance ability.

The COP trajectory data were normalized (WTP to body height and foot length, MML to foot width, and MAP to body height and foot length) and averaged for statistical analysis. The FR was normalized by body height.

One way analysis of variance was used to compare the difference between normal and hemiplegic subjects. Repeated-measure analysis of variance was used to examine the target location effects. Pearson correlation coefficients were used to examine the performance of WBR for targets at various locations. The statistical significant level was set at α = .05 and all statistical analysis was performed using SPSS 8.0 software package for Windows.

RESULTS AND DISCUSSION
As shown in table 1, FR was different between normal and hemiplegic subjects indicating that baseline balance ability of hemiplegic patients was inferior to the balance control of normal subjects. The significant difference in COP trajectory between groups (Table 1) suggested that control of COP during whole body reaching could be an indicator of level of balance. Descriptive analysis showed that the amount of WTP, MML and MAP was larger in hemiplegic patients than in normal subjects, indicating that hemiplegic patients was not able to shift their COP according to the target location as much as normal subjects did. Targets locations were found to imposed graded dynamic balance challenge for both group [3].

The correlation coefficients between FR and COP trajectory were negative, significant and moderate, indicating that the subjects with shorter FR shifted COP less in both frontal and sagittal directions than subjects with longer FR distance [4]. The correlations between FR and COP trajectory in frontal direction were the lowest indicating that FR might not be able to measure the balance control in frontal direction.

CONCLUSIONS
The results of this study suggested that analysis of COP trajectory during WBR can distinguish subjects with different level of balance ability as measured by FR, indicating that WBR could be a dynamic balance training and evaluation tool for hemiplegic patients. Questioning of FR in measuring balance control in frontal direction [4] was supported by the correlation analysis in this study. Analysis of muscle activation patterns might be valuable for interpretation of balance mechanism.

REFERENCES

Table 1. Comparison of group differences.

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Table 2: Pearson correlation coefficients between FR and parameters during whole body reaching.

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THE EFFECTS OF ANTERIOR AND POSTERIOR ANKLE-FOOT-ORTHOSIS ON POSTURAL STABILITY IN HEMIPLEGIC PATIENTS

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INTRODUCTION

Alterations of postural stability are common among patients with hemiplegia following stroke [1]. The impairments of postural stability in poststroke patients include an increase of postural sway during quiet stance [1] and limited weight transfer in both frontal and sagittal planes. Various designs of ankle-foot-orthoses (AFOs) have been found to be able to resolve those impairments but the rationale to choose among different types of AFOs is not yet formulated. The purpose of this study is to compare the effects of different types of AFOs, anterior AFOs vs. posterior AFOs, on static and dynamic standing balance in patients following stroke.

METHODS

The postural stability of 15 patients post 1st and single stroke was measured under the following experiment conditions: (1) quiet stance with shoulder width standing posture and with tandem standing posture; (2) dynamic stance with maximum voluntary weight shifts in anterior-posterior (AP) and medial-lateral (ML) directions. Center of pressure (CoP) trajectory was measured while standing on the Rsscan pressure mat with regular shoes on, with shoes and anterior AFOs (AAFO) on, and with shoes and posterior AFO (PAFO). There were a total of 12 experiment conditions. The parameters representing standing balance were total CoP excursion and maximum length of CoP excursion in AP and ML directions (normalized to body height). Repeated measure analysis of variance was used to compare the effects of different footwear conditions on body sway and maximum voluntary weight shifts. The SPSS 10.0 software package for window was used and the statistical significance level was set at the level of $p < .05$.

RESULTS AND DISCUSSION

Non-significant footwear and standing posture interaction effects were found (Table 1, $p = .186$). The main effects of footwear conditions was non-significant (Table 1, $p = .072$). The post hoc pairwise comparison indicated that no significant difference between them (Figure 1, $p = .39$, $p = .11$, $p = .1$). The main effects of standing posture was significant (Table 1, $p = .001$), indicating that the body sway was different among experiment conditions no matter which footwear conditions the patients were at. The post hoc pairwise comparison showed that tandem stance induced significantly more body sway in stroke patients no matter which footwear conditions the patients were under (Figure 1). The main effects of footwear in ML direction weight shifting was significant ($p = .014$), and in AP direction weight shifting was no significant ($p = .905$), indicating that the maximum voluntary weight shifts in ML direction was different among footwear conditions. The post hoc pairwise comparison showed that PAFO improved significantly more weight shifting ability in ML direction. Narrowing the BoS increased the postural demands for the stroke patients.

As shown in Table 1 and Figure 1, although the footwear conditions failed to affect standing balance significantly, possibly due to the small sample size in this study, the patients wearing the AFOs and PAFO tended to improve their maximum voluntary weight shifts in ML direction and body sway during quiet stance. The effects of decreased body sway were similar between AAFO and PAFO

<table>
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<td>3.69</td>
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*p < .05; FW: footwear, SP: standing posture

CONCLUSIONS

The different types of AFOs showed significant different effects on weight shift ability in ML direction, but no different effects on body sway. The PAFO affected weight shifts ability in ML direction more effectively than the AAFO did.

REFERENCES

A STUDY OF DUAL TASK EFFECTS ON POSTURAL STABILITY IN STROKE PATIENTS

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Introduction

The dual task paradigm is an often encountered situation in daily living[1]. The competition of attentional demands of both the postural and suprapostural task could influence the performance of either task[2]. After brain injury, especially in stroke patients, the cognitive and motor performance may be quite different from healthy people[3]. Understanding the interference effects of dual task and postural task is necessary for training program design. The purpose of the study was to investigate the interference that occurs between concurrent performance of postural task and the visuospatial tasks. Dependent variables in postural stability were: (1) medial-lateral center of pressure (CoP) trajectory pathway, (2) anterior-posterior CoP trajectory pathway, (3) CoP trajectory total pathway. Parameters in supraspatial tasks were mean reaction time on each task condition.

Materials and Methods

Nineteen subjects post 1st stroke onset were recruited. All subjects underwent both the single task and dual task experiment conditions. Subjects performed all trials on a 1m instrumented mat (Footscan, RSscan pressure measurement system, Belgium) and the sampling frequency was 20 Hz. In single task conditions, the postural stability of the stroke patients were measured while standing on the RSscan pressure mat with the following three stance configurations (a) shoulder-width double stance, (b) step stance with the sound leg at the back, (c) step stance with the affected leg at the back. In dual task conditions, the postural stability was measured while stance as in single task conditions and reacted to a suprapostural conditions which contained visuospatial tasks, with a two-button answer switch on the sound lateral thigh. And there are two levels (easy & complex) in the visuospatial task. Figure 1& 2

Repeate-sexes measures two-way analysis of variance was used to examine the effects of dual tasks on postural stability performance with significant level setting at α =.05.

Table 1 ANOVA on single and dual task conditions

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*p<0.01  **p<0.05

Results and Discussions

Significant postural and visuospatial tasks interaction on CoP_total was found (Table 1, p=0.02). Significant simple main effects on postural task were found. There are no significant posture and visuospatial task interaction on CoP_AP & CoP_ML was found (Table 1, p=0.63, 0.37). Significant main effects on postural task were both found (Table 1, p<0.01). The effect of visuospatial tasks on the mean reaction time were not significant both visuospatial tasks (p>0.05).

Postural stability during visuospatial task conditions was influenced by different posture conditions. The dual task conditions decreased the postural sway on medial-lateral, anterior-posterior, and CoP total pathway as compared to the single task conditions[2,3], especially when the postural tasks were hard and the challenges of postural control was high[2].

Conclusions

Dual task conditions decreased the postural sway, that is, stroke patients might tend to cocontract their muscles to maintain postural stability in high-demanding suprapostural tasks.

References

**FINITE ELEMENT STRESS ANALYSIS OF THE INTERvertebral DISC IN A RHEUMATOID ARTHRITIS PATIENT WITH THE CERVICAL INVOLVEMENT**


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Introduction

Rheumatoid arthritis (RA) often causes deformation and abnormal kinematics of the cervical spine. The cervical deformation and abnormal kinematics associated with RA may affect stress distribution of the intervertebral disc (IVD) in RA patients. Recent advancement of medical imaging techniques allow three dimensional (3D) in vivo measurement of segmental movement of the spine [1,2]. A kinematic analysis of the cervical spine was conducted in RA patients and demonstrated abnormal kinematics associated with RA [3]. The purpose of this study was to establish a finite element stress analysis method of cervical IVD in a RA patient using a 3D computer model reconstructed from computer tomography (CT) and magnetic resonance image (MRI) and using kinematic data of the cervical spine.

Methods

A RA patient with deformation of vertical subluxation (VS) and subaxial subluxation (SS) underwent CT scans in a neutral position and MRI scans in three different positions in supine position: neutral, flexion, and extension. Five IVD models from C2/3 to C6/7 discs in each position were created utilizing the inferior and superior endplate surface geometrical data obtained from the 3D CT model. Since MRI of the IVD at each level demonstrated mild or severe IVD degeneration, Young’s modulus and Poisson’s ratio of annulus fibrosus (AF) and nucleus pulposus (NP) were set as 4.7 MPa and 0.45, respectively [4]. Transformations of each endplate in flexion and extension positions were determined using a volume merge method [2], and these results were applied to the superior nodes of IVD as forced displacements. The inferior nodes of IVD were fixed in all directions.

Results and Discussion

Maximum compressive stresses were noted in the right sides of the IVD and maximum tensile stresses were noted in the left sides of the IVD at C2/3 and C3/4 levels in flexion (Figure 1). In extension, however, maximum compressive stresses were noted in the left side of the IVD and maximum tensile stresses were noted in the right side of the IVD (Figure 2). These results seem to be caused by bending segmental motion at C2/3 and C3/4 during flexion and extension, which may be caused by asymmetric destruction of the facet joints associated with RA. The stresses calculated at C5/6 level in flexion was predominantly tensile. This may be due to a paradoxical movement caused by the method to provide the flexed position, which might have caused neck destruction. An MRI examination in standing position will be needed to study cervical kinematics and stress analysis of the cervical IVD during activities of daily living.

Acknowledgment

This study was supported by the Academic Frontier Research Project on " New Frontier of Biomedical Engineering Research " of Doshisha University & Ministry of Education, Culture, Sports, Science and Technology.

References

INFLUENCE OF STEM DESIGN ON CONTACT PRESSURE AND STRESS DISTRIBUTION IN REVISION TOTAL KNEE REPLACEMENT

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Introduction

Long stems are frequently used in revision total knee replacement (TKR) both to provide additional fixation and to assist in ensuring more consistent component alignment. Many clinical studies have supported that the implant with longer stem length provides better stability and less micro motion. However, it has been reported that a press-fit stem in revision TKR could induce the stem-end pain [1].

In this study, we investigated the influence of stem-end design on contact pressure and stress distribution in revision since it is clinically hypothesized that the stem-end pain would be related to the local contact pressure and stress distribution around the stem-end.

Materials and Methods

A finite element model of a tibia was created from 1mm-CT scan images and implanted with a commercial tibial prosthesis (Wright Medical Technologies, U.S.A.) by using commercial pre/post processor FEMAP® (V8.2, EDS Corp., U.S.A). The elastic modulus and Poisson’s ratio of cortical and cancellous bone were respectively 17GPa, 0.36 and 300MPa, 0.3. For the tibial prosthesis, the elastic modulus and Poisson’s ratio, which was Titanium, were 110GPa and 0.3.

Applied load to the tibia was 2000N which represented the peak loading based on the gait cycle and distributed 60% to the medial tray and 40% to the lateral tray [2]. Cemented tray and press-fit stem, which represented the hybrid fixation, were applied. The finite element analysis were carried out using ABAQUS™ (Standard 6.5, ABAQUS Inc., U.S.A). In order to investigate the contact pressure and stress in the stem-end section, various 3D CAD models were developed and four design parameters which were length, diameter, slot and press-fit were applied in the pre-operative planning.

Results

High contact pressure and von Mises stress were occurred near the stem-end regardless of the change of stem design parameters and the von Mises stresses were especially higher in the medial region.

As the stem length decreased, the peak contact pressure and von Mises stress were reduced: for 115 mm, 90 mm, and 70 mm of stem length, the contact pressures were 7.13 MPa, 1.60 MPa, and 1.49 MPa respectively as the change of stem design parameters and the von Mises stresses in the medial region were 5.80 MPa, 0.16 MPa, and 0.18 MPa respectively. Both the contact stress and von Mises stress were dramatically reduced when the length was changed from 115 mm to 90 mm. The peak contact pressure and von Mises stress were increased as the change of stem diameter from 10 mm to 12 mm: the contact pressure increased by 16% while the peak von Mises stresses in the medial region increased by 34%.

As the press-fit magnitude increased, the peak contact pressure was substantially increased from 7.13 MPa to 20.96 MPa and 43.06 MPa for 5 µm, and 10 µm, respectively. Similarly, the peak von Mises stresses in both the medial and the lateral regions increased. The slot shape at the stem-end influenced only on the contact pressure. The peak contact pressure in the model with the slot was reduced to 2.40 MPa from 7.13 MPa while the von Mises stresses were similar in spite of the change of slot shape.

Discussion

There is a hypothesis that peak contact pressure and stress are related to the stem-end pain [3]. In the clinical study, the incidence of stem-end pain was reported in the group using longer stems. In the case of using thicker stems, there was not a significant difference to feel the pain. However, a group with the stem-end slot responded the significant reduction of the stem-end pain [1]. Compared with those clinical reports, the results of this study were coincident with the above hypothesis.

Therefore, the results of this study demonstrated that the shorter stem length, smaller stem diameter, and slotted design was planned, the smaller peak contact pressure and stress were induced in the bone adjacent to the stem-end and could reduce the stem-end pain. When a press-fit stem was planned, the peak contact pressure and stress increased drastically and it was represented that large clearance press-fit could induce the stem-end pain. The result of this study could be useful information for not only the pre-operative planning but also the implant design modification in revision TKR.

References


Acknowledgement

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Nonlinear buckling analysis for etiology study of idiopathic scoliosis

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Introduction

Idiopathic scoliosis still remains in disorders of unknown etiology. The authors presented a hypothesis that the etiology of the idiopathic scoliosis is a buckling phenomenon induced by the growth of vertebral bodies based on the accordance of the clinical single and double-major curves with the results of the fourth and sixth buckling modes respectively analyzed using a finite-element spine model by the linear buckling theory [1]. In their recent study, deformation paths induced by the growth of vertebral bodies were analyzed conscientiously by the finite-element method considering geometrical nonlinearity along the growth. In these analyses, to prevent the generation of the controllable modes by posture change, an additional boundary condition around the node of the fourth buckling mode for the single curve was used in which movement was constrained in the transverse plane and remained in vertical direction. In three cases fixing C6, C7 and T1, deformation paths through unstable limit points up to stable points including side bending modes corresponding to the clinical single curve were obtained.

However, other complex movements such as rebounding to the opposite side were included. From the observation, it was suspected that the calculated paths were affected by the boundary conditions. In this study, the deformation path is reanalyzed using a weaken boundary condition supporting the cervical vertebra with distributed springs.

Method

The spine finite element model constructed in the previous study without rib cage was used. The growth of vertebral bodies was modeled by the non-elastic bulk strain occurring uniformly in epiphyseal rings and hyaline cartilage plates. The growth was assumed to occur in the vertebral bodies from T4 to T10, in which case the buckling of thoracic single curve is easy to occur [2], with proportion of the bulk strain $\alpha$ shown in Table 1.

The boundary conditions were assumed as follows. The sacrum was fixed as the base position of deformation. To prevent the generation of the controllable modes by posture change, we assumed that the open boundaries of C2 to C7 were supported with springs uniformly distributed in the normal direction within the transverse plane remaining the vertical movement. The spring factor was decided to deform with 1/10 of the displacement at loading the front top of C2 in the side direction. During the growth deformation, it was assumed that the directions of the springs keep the initial directions.

For nonlinear buckling analyses induced by initial strain, we developed a program using the finite element method. To follow up the deformation path after unstable limit points, we used the arc-length method. To judge the buckling points, we employed the method using $LDL^T$ decomposition of the tangent stiffness matrix [3]. Using the method, we monitor the number of the negative components in $D$ matrix and find passing the buckling points by the change of the number.

Table 1: Proportion of growth strain $\alpha$

<table>
<thead>
<tr>
<th>Vertebra</th>
<th>$U_1$</th>
<th>$U_2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>T4</td>
<td>0.05</td>
<td>0.05</td>
</tr>
<tr>
<td>T5</td>
<td>0.10</td>
<td>0.10</td>
</tr>
<tr>
<td>T6</td>
<td>0.125</td>
<td>0.125</td>
</tr>
<tr>
<td>T7</td>
<td>0.15</td>
<td>0.15</td>
</tr>
<tr>
<td>T8</td>
<td>0.10</td>
<td>0.10</td>
</tr>
<tr>
<td>T9</td>
<td>0.05</td>
<td>0.05</td>
</tr>
<tr>
<td>T10</td>
<td>0.05</td>
<td>0.05</td>
</tr>
</tbody>
</table>

Results and Consideration

The result of the deformation path at the front-center point on T8 is shown in Fig. 1. The deformations at P1, P2, P4 and P5 were forward or backward single bending modes in the sagittal plane. That at P3 was a side bending mode corresponding to the clinical single curve. In this result, no path rebounding to the opposite side was observed. Based on the result, we can consider the followings.

1. The rebounding path to the opposite side is caused by the overconstraint of the upper boundary condition.
2. The forward bending at P1 represents the decreasing of the thoracic kyphosis during the growth spurt observed for normal children.
3. The side bending mode at P3 occurs stably when the growth rate reaches 0.645, that is under the $P_3$, and jumps in the mode at P3 by some exercise.
4. The maximum deflection of 0.65[mm] at $P_3$ is too small for the etiology of idiopathic scoliosis. We need to consider the remodeling by the stress at $P_3$.

References

Experimental Consideration in the Arch of the Foot by Stress Freezing Method

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Introduction

As for foot joint, especially complex shape is had in the load joint that supports weight, and there are a lot of indistinct points in the mechanism. Recently, the number of foot joint syndromes by an acquired transformation increases. Because loading condition and the structure of foot joint greatly influence them, the clarification of the mechanical property of foot joint is important. Especially, a flatfoot has a lot of patients as an acquired case. Not only a medical viewpoint but the report of research from an engineering viewpoint is called for[1][2]. In this research, the bone in the arch of a normal foot and a flat foot was reproduced by three-dimensional model, the stress distribution was analyzed by the photoelastic investigation, and it made comparative study of them.

Materials and Methods

The material used in this research is the epoxy resin which mixed ALALDAITE (CT-200) and a hardening agent (HARDNER HT-901) at a rate of a bulk density 100:30. The material characteristics of an epoxy resin are shown in Table 1.

Table 1 Material Characteristics of Epoxy resin

<table>
<thead>
<tr>
<th>Condition</th>
<th>Young’ Modulus E [MPa]</th>
<th>Poisson’ Ratio ν</th>
<th>Photelastic Sensitivity α [mm/N]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Room Temperature 2940</td>
<td>0.30</td>
<td>0.10</td>
<td></td>
</tr>
<tr>
<td>120~130°C</td>
<td>13.62</td>
<td>0.48</td>
<td>4.00</td>
</tr>
</tbody>
</table>

The examination model produced the mold of a left foot from the foot joint model of a medicine model, and cast each bone in the epoxy resin, respectively. Then, a bone model is constructed with elastic adhesives. It referred to the Yokokura method which is a method of judging flat-footed the foot arch fell, and the bony structure of foot joint greatly influence them, the clarification of the mechanical property of foot joint is important. Especially, a flatfoot has a lot of patients as an acquired case. Not only a medical viewpoint but the report of research from an engineering viewpoint is called for[1][2]. In this research, the bone in the arch of a normal foot and a flat foot was reproduced by three-dimensional model, the stress distribution was analyzed by the photoelastic investigation, and it made comparative study of them.

Results and Discussion

Consideration is advanced about Calcaneus as which the remarkable difference was regarded in the analyzed bone. The striped photograph and stress distribution map of Calcaneus-1 of a healthy state are shown in Figure 1.

Stress distribution of Fig. 2 resembles Fig. 1. However, in the flat state, the maximum stress occurred in the slice 3. This is shown in Figure 2.

Conclusions

About the normal foot and flatfoot of a foot joint dynamics model, the following knowledge was acquired from the experiment analysis by a stress freezing method. If a foot becomes flat, the stress of sidewall will change outside from an inner side. As a result, it is thought that change arose in stress distribution of Calcaneus.

References

AUTOMATED IMAGE MATCHING FOR 3-D POSE ESTIMATION OF KNEE USING SINGLE-PLANE FLUOROSCOPY


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Materials and Methods

True value of relative pose: The study was carried out on three fresh porcine cadaver knees. Three sphere markers were bonded to each femur and tibia. The femur and tibia were fixed using an external fixation device, resembling an anatomical position. The central coordinates of the sphere markers were measured by a 3D coordinate measuring machine (BH504, MITUTOYO, Japan) to define the absolute position of each bone. The relative pose between the two bones was derived from their absolute positions, and used as the true value.

DRR creation using bone volume model: Three dimensional bone volume models were generated by bone segmentation of CT scan data of femur and tibia (field of view: 640 x 512 pixel, pixel size: 0.35 x 0.35 mm², slice thickness: 1 mm). The original anisotropic voxel, 0.35 x 0.35 x 1.0 mm³, was transformed into an isotropic voxel with 1.0 x 1.0 x 1.0 mm³ to improve the quality of DRR. DRR was created by projecting the all voxels to the image plane by the following equation.

\[ I_c(u,v) = \sum_{i=1}^{n} \alpha_i I_0 \]

Where \( I_0 \) indicates the dynamic range, \( n \) the number of voxels projected to a point \((u,v)\) on the image plane, \( \alpha \) the attenuation factor, \( \mu \) the voxel value. Values of \( I_0 \) and \( \alpha \) were selected so that the value of \( I \) matched with that of the actual fluoroscopic image.

Automated pose estimation algorithm: To estimate the 6 DOF parameters of bone, a gradient difference of fluoroscopic image and DRR [1] was used as a similarity measure. Maximizing the similarity measure between the 2 images was carried out using a downhill simplex algorithm [2]. To compare the relative pose of femoral model with respect to tibial model, 6 DOF parameters of femoral model was first determined with a manual image matching [3]. Then the femoral model was placed according to the true value of relative pose. Finally, the automated image matching algorithm was performed on the femoral model. The estimated relative pose with respect to the tibial model was compared with the true value of relative pose. Ten single plane fluoroscopic images with different flexion angles and different directions of image acquisition were studied.

Results and Discussion

Results of the absolute errors in estimating 6 DOF parameters are listed in Table 1. Errors in two out-of-plane (y and z) rotations, 2.8 deg and 1.7 deg, and one out-of-plane (x) translation, 3.9 mm are larger than those in in-plane (x) rotation, 0.7 deg, and two in-plane (y and z) translations, 0.5 mm and 0.7 mm. These accuracies are comparable to previous report [4]. Ideally, errors could be negligible since the automated image matching started from the true value. However, several factors such as less clear bone edge due to surrounding soft tissue and imperfect bone volume models would make the accuracy worse. Therefore, although the present result shows a potential of the proposed method for the accurate analysis of in vivo knee kinematics, further research is required to identify sources of errors responsible for inaccuracy.

Table 1 Mean (SD) of absolute errors in estimating rotation and translation parameters (n=10)

<table>
<thead>
<tr>
<th>Rotation</th>
<th>Translation</th>
</tr>
</thead>
<tbody>
<tr>
<td>x,deg</td>
<td>y,deg</td>
</tr>
<tr>
<td>x,mm</td>
<td>y,mm</td>
</tr>
<tr>
<td>0.7</td>
<td>2.8</td>
</tr>
<tr>
<td>3.9</td>
<td>0.5</td>
</tr>
<tr>
<td>(0.4)</td>
<td>(2.3)</td>
</tr>
</tbody>
</table>

Conclusions

A method for analyzing in vivo knee kinematics using single plane fluoroscopy and 3D bone models was developed. Digitally reconstructed radiographs (DRRs) were generated from 3D bone volume models by a voxel projection technique. The relative 3D-pose (full 6 DOF parameters) between porcine femur and tibia were determined by matching the DRR of each bone model with the fluoroscopic image by maximizing the similarity measure between the 2 images. The true value of the relative pose was measured by a 3D coordinate measuring machine. The mean errors of the two out-of-plane rotation parameters were 2.8 deg and 1.7 deg, and the in-plane parameter was 0.7 deg. The error of out-of-plane translation parameter was 3.9 mm while the two out-of-plane parameters were 0.5 mm and 0.7 mm.

References

EFFECTS OF ANTERIOR MALLEAR LIGAMENT FIXATION ON HEARING IN HUMAN MIDDLE EAR MODEL

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Complete bony fixation of the malleus is a well documented disorder after stapes surgery. A comprehensive numerical model of human middle ear can provide better understanding of the possible causes of unfavorable results. In this study, a three-dimensional finite element model of human middle ear is proposed that includes tympanic membrane (TM), ossicular bones, middle ear suspensory ligaments/muscles, and inner ear fluid. This model is constructed based on a complete set of CT images of a volunteer (right ear). The validity of this model is confirmed by comparing the movements of tympanic membrane and stapes footplate obtained by this model with published experimental data. Analysis of the dynamic behavior of human middle ear is performed by the FE model for fixation of anterior mallear ligament (AML). Similar, significant changes of malleus vibration patterns for AML fixation are found by FEM calculations. The final FE model is shown to be reasonable in predicting hearing loss of AML fixation.
BIOMECHANICAL ANALYSIS OF WRIST JOINT LOADING BETWEEN OPEN AND CLOSE KINETIC CHAIN EXERCISE

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INTRODUCTION
Various strengthening program, such as open-chain and close-chain kinetic exercises, are used for sports activity and post-traumatic rehabilitation. Nonetheless, there is very little data available regarding to the benefits and drawbacks between the two types of exercises for the upper extremity. The purpose of this study is to investigate the effect of different kinetic chain exercises on wrist joint loading.

MATERIALS AND METHODS
Ten male subjects volunteered in this study. Their average age was 25 years, with an average height of 174 cm, and average weight of 67.3 Kg. With both hands in neutral position, each subject was asked to perform push-up (close-chain) exercise and bench press (open-chain) exercise. A Motion Analysis System (Motion Analysis Inc., Santa Rosa, CA, U.S.A.), two Kistler force plates (Model 9281B, Kistler Instruments AG, Winterthur, Switzerland), and computers were used to collect data.

RESULTS
In the open-chain exercise, the medial-lateral shearing forces around the wrist joint is 2 times higher than the close-chain exercise; and the radial-ulnar and the flexion-extension moment were 1.8 times and 1.6 times higher than the close-chain exercise, respectively.

DISCUSSION AND CONCLUSIONS
The loading biomechanics of the wrist joint differs with open and close kinetic chain exercises. There were more medial-lateral shearing forces and flexion-extension moments around the wrist joint during the open-chain exercise, which could be more harmful in the earlier stage of the rehabilitation. In conclusion, these data will be valuable reference in clinical treatment and rehabilitation.
Introduction: Walking with high-heel shoes can make the young females look elegant, but thus easily does harm to their feet and lower limbs. The purpose of this paper is to make gait analysis on ten females’ walking wearing different heel height shoes and provide some scientifically theoretical basis in quantity for studying the effect of heel heights on the gait of young females and the design of the high-heel shoes.

Materials and Methods: In this paper, normal digital video camera and footscan® USB system are employed and make comparative analysis on the kinematic parameters and plantar pressure parameters by ShiXun motion analytic system and F-scan Plate System. when 10 young females (average age is $19.2 \pm 1.0$) walk on the flat ground with normal speed with barefoot and wearing four pairs of shoes with varying heel heights. The runway is covered with materials EVA for protecting it and making Footscan® USB Plate not influence the subjects walking normally. We use SPSS11.5 for Data processing and make statistic analysis for the data. All data are expressed by $X \pm SD$. The differences of the interrelated Parameters are tested through Ttest.

Results and Discussion: When the subjects walk wearing the high-heel shoes, 1) the pace length is shorter, the pace velocity is slower and the gait cycle is longer. Single support phase occupies smaller percentage and double support phase occupies bigger percentage, in order to hold balance and stability. 2) Ankle joint, knee joint and hip joint move inflexible and the motion range of these joints grow narrower, which make the muscle of rear group shorter and render the cushion of the lower limb to reduce. The result will perhaps increase the possibility of the lower limb fatigue and even injury. 3) The track excursion of the center of plantar pressure became bigger, keeping balance become harder. 4) The peak force of forefoot, the percentage of impulse on the forefoot and mid-foot increase, which cause the foot fatigue, pain and injuries easily or the abnormality of foot growth. 5) In addition, the thickness of shoes heel also influences the gait characteristic.

Conclusions: Walking with the high-heel shoes has great damage on foot growth of young female and will possibly influence the knee joint and lumbar, so we advice young female not to wear high-heel shoes often.

References:
Muscle reflection human model in wheelchairs traveling in motor vehicles supported by a seatbelt and headrest

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Introduction

Recently, a lot of disabled people feel endangered when they travel in the vehicle sitting on a wheelchair. But no safety standard is applied to the in-vehicle wheelchair during driving. According to the questionnaire of the wheelchair user in a motor vehicle, disabled people don't attach seatbelt and headrest to make safety guideline which is acceptable for wheelchair user as soon as possible. The purpose of this research is to make a muscle reflection human model supported by a seatbelt and headrest to validate these fixation apparatus.

Rigid link model

Rigid link model in this study is shown Fig.1

Human body sitting on the wheelchair is expressed by a link segment model of three degrees of freedom consist of head, hips and human trunk. Motion of this model is in two center of rigid body. Based on these assumption, dynamic equation of Lagrange is given by eq(1)

\[
\begin{bmatrix}
    m_1 + m_2 + m_3 & m_2 r_1 + m_2 r_2 + m_2 I_1 + m_2 I_2 & m_2 I_2 \\
    m_2 r_1 + m_2 r_2 + m_2 I_1 + m_2 I_2 & m_2 I_1 + m_2 I_2 + m_2 I_2 \\
    m_2 r_1 & m_2 r_2 + m_2 I_2 & m_2 I_2 + m_2 I_2 \\
\end{bmatrix}
\begin{bmatrix}
    \ddot{\theta}_1 \\
    \ddot{\theta}_2 \\
    \ddot{\theta}_3 \\
\end{bmatrix}
= \begin{bmatrix}
    0 \\
    0 \\
    0 \\
\end{bmatrix}
+ \begin{bmatrix}
    B_i \\
    F_i \\
    M_i \\
\end{bmatrix}
\]

Input is surface acceleration \( \ddot{x} \). Output is horizontal displacement of lumbar spine \( X \), angle of cervical and trunk part \( \theta_1, \theta_2, \theta_3 \), \( m_j \) is the mass of each part, \( L_i \) is the length of link, \( r_i \) is the length from center of gravity to end, \( I_i \) is the inertia moment, \( F_i \) is frictional force of seating force.

\( B_i \) is the tension of a seatbelt, \( N_i \) is the reaction force by the back rest and headrest, \( M_i \) is the torque of joint. To calculate \( M_r \), torque of active and passive muscle was separated. \( T_a \) is defined as an active muscle torque, \( T_p \) is defined as a passive muscle torque.

Experiment

There are two purposes of this experiment. One is measurement of the behavior of disabled people and another is comparison between experiment and simulation. Carriage experiment device is shown Fig.2

Results and Discussion

Some important data about behavior of disabled people was obtained in this study. The simulation model is not only practical as a computational research tool, but also significant as a safety assessment tool of fixation apparatus.

References

Impact Analysis of Human Limb Muscle under Head, Wrist and Thrust Attack of Kendo
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Introduction
Kendo is one of the popular sports in modern life. Head, wrist and thrust attack are the efficient skill to get winning point during match. Human muscle skeletal model was developed for biomechanical study. The human model was consists with 19 bone-skeleton and 122 muscles. Muscle number of upper limb, trunk and lower limb part were 28, 60, 34 respectively. Bone was modeled with 3D beam element and muscle was modeled with spar element. The biomechanical stress and strain analysis of human muscle was conducted by proposed human bone-muscle finite element analysis model under head, wrist and thrust attack for kendo.

Analysis of muscle-skeleton model and human behavior
For the human impact structural analysis, the human model of kendo head, wrist and thrust attack is developed. Human with 180 cm height, 66 kg weight was modeled and the bamboo knife of 120 cm length, 0.5 kg weight was applied. The human model consists of 19 segments and 18 joints. The elastic modulus of the skeletal model, Poisson’s ratio and mass density are 20 GPa, 0.3 and 2000 kg/m³ respectively. Total number of developed muscles was 122. The elastic modulus, mass density and initial strain of muscles were 40 MPa, 1500 kg/m³ and 0 respectively. Total number of developed muscles was 122. The human model was consists with 19 bone-skeletal and 122 muscles. Muscle number of upper limb, trunk and lower limb part were 28, 60, 34 respectively. Bone was modeled with 3D beam element and muscle was modeled with spar element. The biomechanical stress and strain analysis of human muscle was conducted by proposed human bone-muscle finite element analysis model under head, wrist and thrust attack for kendo.

Results and Discussion
The stress and strain analysis of human limb muscle under kendo head, wrist and thrust attack was conducted on the basis of a finite element human model. In the case of head, wrist and thrust attack, deformed posture after attack returns to initial posture after 1.1 s, 1.1 s and 1.3 seconds respectively. Maximum stress was about 12.4 MPa occurred on the bamboo knife and hand grip at head attack motion. Figure 1 (a), (b), (c), (d) show the stress distributions of the muscle-skeleton model at 0.56 seconds and 0.77 seconds. At 0.56 seconds of head and wrist attacks, the maximum displacements of the end of a bamboo knife were 61.5 mm and 74.4 mm. At 0.77 seconds of thrust attack, the maximum displacement of the end of a bamboo knife is 23.4 mm.

For the human impact structural analysis, the human model of head attack was carried out by 1.1 seconds and 1.3 seconds. Figure 1 (a) shows the stress distributions of the head hitting posture at 0.56 seconds. Maximum stress was about 12.4 MPa occurred on the bamboo knife and hand grip at head attack motion. Figure 1 (b), (c), (d) show the displacement distributions of the muscle-skeleton model at 0.56 seconds and 0.77 seconds. At 0.56 seconds of head and wrist attacks, the maximum displacements of the end of a bamboo knife were 61.5 mm and 74.4 mm. At 0.77 seconds of thrust attack, the maximum displacement of the end of a bamboo knife is 23.4 mm.

Figure 2 shows the strain responses of deltoid, triceps brachii, biceps brachii and extensor carpi radialis longus muscles of human body. At 1.1 seconds after head hitting, the strain response recovered to the initial state (Figure 2 (a), (b)). Deltoid, biceps brachii and triceps brachii muscle showed almost negative % strain and appeared two salient contractions and relaxations after head hitting. Extensor carpi radialis longus muscle had three contraction and relaxation periods.

References
The Effect of Carbon and Glass fiber Material of Vaulting Poles on the Vaulter’s Techniques
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Introduction

Pole vault basically is the process of energy transformation; it refers to the changes between kinetic and potential energies (including the potential energy of human gravity and pole’s flex level). Nowadays, the “glass fiber” materials vaulting pole has been widely used by pole vaulters in the world. This type of material allows vaulter to transfer kinetic energy into maximum potential energy and store in the pole by having efficient acceleration during run up. Moreover, the vaulter create a good swing around the shoulder right after take-off and the drive of torso to well cover the pole before the pole start to recoil, in order to pop the vaulter over the bar successfully. The main purpose of this study was to discuss the technical differences of vaulters’ who used carbon and glass fiber poles.

Materials and Methods

There were six elite male Taiwanese pole vaulters who served as the subjects and their personal information were as followed:

Table 1 Basic data of subjects

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Height</th>
<th>Weight</th>
<th>Best record</th>
<th>Experience</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>26</td>
<td>173</td>
<td>70</td>
<td>5m10</td>
<td>12</td>
</tr>
<tr>
<td>B</td>
<td>28</td>
<td>190</td>
<td>78</td>
<td>4m80</td>
<td>3</td>
</tr>
<tr>
<td>C</td>
<td>21</td>
<td>186</td>
<td>79</td>
<td>5m00</td>
<td>5</td>
</tr>
<tr>
<td>D</td>
<td>24</td>
<td>172</td>
<td>62</td>
<td>4m90</td>
<td>10</td>
</tr>
<tr>
<td>E</td>
<td>23</td>
<td>173</td>
<td>63</td>
<td>5m01</td>
<td>8</td>
</tr>
<tr>
<td>F</td>
<td>21</td>
<td>168</td>
<td>70</td>
<td>4m50</td>
<td>3</td>
</tr>
</tbody>
</table>

Two high speed camcorders(60Hz) were used in the experiment to monitor the kinestatic parameters of each vaulter’s take-off. One was placed from the left side of the take-off point, the other one was located a little bit behind the first camcorder(shown as figure 1). The Arial Performance Analysis System 3D software, SPSS 10.0 and Microsoft Office Excel were applied to do Graphic and statistical analysis.

Results and Discussion

Comparing these two different pole materials, there was statistical significance on vaulters’ left leg swing(see Table 2). When using carbon fiber pole, vaulter showed slower swing speed since the carbon fiber pole was harder to penetrate comparing with glass fiber pole. Kukureka’s study showed that pole vaulter needed to have good and fast leg swing ability to conquer carbon fiber pole. [1]

Table 2 The duration and maximum angular velocity of leg swing

<table>
<thead>
<tr>
<th></th>
<th>Mean±SD</th>
<th>t</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Duration of leg swing (s)</td>
<td>0.017±0.023</td>
<td>-1.746</td>
<td>0.141</td>
</tr>
<tr>
<td>Maximum angular velocity of leg swing (%/s)</td>
<td>66.31±22.29</td>
<td>7.287</td>
<td>0.001*</td>
</tr>
</tbody>
</table>

* p < .05

As table 3 showed, carbon fiber pole’s maximum bend angle was far less than that of glass fiber pole(t=2.892, p=0.034) due to its material characteristic [1].

The glass fiber pole penetrated faster than carbon fiber pole, vaulter needed to complete the take-off and swing techniques fast enough before the pole reached the horizontal plane. On the other hand, the carbon fiber pole had faster recoil speed which would bring the vaulter’s body mass higher and help them to fly over the bar with less difficulties.

Table 3 The difference of glass and carbon fiber when poling

<table>
<thead>
<tr>
<th></th>
<th>Mean±SD</th>
<th>t</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum pole bend angle</td>
<td>6.56±5.56</td>
<td>-2.892</td>
<td>0.034*</td>
</tr>
<tr>
<td>The average angle velocity of pole</td>
<td>1.97±4.52</td>
<td>-1.068</td>
<td>0.334</td>
</tr>
<tr>
<td>The average angle speed between pole penetration and recoil</td>
<td>-2.05±5.19</td>
<td>0.966</td>
<td>0.379</td>
</tr>
<tr>
<td>The average angle speed of pole recoil</td>
<td>0.27±1.18</td>
<td>-.555</td>
<td>0.603</td>
</tr>
<tr>
<td>The angle of body mass during pole recoil phase</td>
<td>-1.26±.90</td>
<td>3.444</td>
<td>0.180</td>
</tr>
</tbody>
</table>

* p < .05

Conclusions

Different pole material would never affect vaulter’s take-off technique. When using carbon fiber pole, the vaulter’s maximum angle speed of swing was slower than that of glass fiber pole. The carbon fiber pole had faster pole recoil speed; hence, vaulter’s body mass angle would basically remain the same which would minimize the energy loss.

References

The Comparison of Flex Performance between New Carbon Fiber and Glass Fiber of Vaulting Poles

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Nan Kai Institute Of Technology, Nantou County, Taiwan

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Introduction

There are three major considerations when designing vaulting pole: (1) Good ability to store potential energy (2) Appropriate pole flex to meet vaulter’s need (3) Lighter pole weight to carry easier and accelerate faster [1]. The study discovered that carbon fiber pole possessed the features of low intensity, high durability and flex. In other word, the carbon fiber pole is called the modern metal pole. The main purpose of this study was to compare the differences of pole bend amount and recoil speed between carbon and glass fiber poles.

Materials and Methods

A 60 Hz camcorder was applied to monitor the amount of static and dynamic pole bend, recoil speed under 10.15.20.30kg weight load. The set up of the experiment was as Figure 1.

The Arial Performance Analysis System 3D software, SPSS 10.0 and Microsoft Office Excel were applied to do Graphic and statistical analysis.

Results and Discussion

There was not statistical significance between static and dynamic pole bend amount. They all showed the same amount of pole bend when loaded under the same weight. Shown as Table 1.

Table 1 Static and dynamic pole bend amount (unit: cm)

<table>
<thead>
<tr>
<th>Load</th>
<th>Glass fiber pole</th>
<th>Carbon fiber pole</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>static</td>
<td>dynamic</td>
</tr>
<tr>
<td>10kgw</td>
<td>6.1</td>
<td>9.06±1.10</td>
</tr>
<tr>
<td>15kgw</td>
<td>9.1</td>
<td>15.07±1.62</td>
</tr>
<tr>
<td>20kgw</td>
<td>12.0</td>
<td>19.14±3.21</td>
</tr>
<tr>
<td>25kgw</td>
<td>14.5</td>
<td>24.03±1.09</td>
</tr>
<tr>
<td>30kgw</td>
<td>18.0</td>
<td>28.84±0.81</td>
</tr>
</tbody>
</table>

When loaded 10 and 15kg, the carbon fiber pole showed faster recoil speed and slower pole bend speed. Shown as Table 2.

Table 2 The difference between two materials' maximum dynamic pole bend speed (unit: m/s)

<table>
<thead>
<tr>
<th>Load</th>
<th>Mean±SD</th>
<th>t</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>10kgw</td>
<td>-13.49±6.333</td>
<td>-2.13</td>
<td>0.059</td>
</tr>
<tr>
<td>15kgw</td>
<td>-2.35±6.34</td>
<td>-0.37</td>
<td>0.719</td>
</tr>
<tr>
<td>20kgw</td>
<td>1.05±3.35</td>
<td>0.31</td>
<td>0.761</td>
</tr>
<tr>
<td>25kgw</td>
<td>-2.30±3.85</td>
<td>-0.60</td>
<td>0.563</td>
</tr>
<tr>
<td>30kgw</td>
<td>-3.11±3.15</td>
<td>-0.98</td>
<td>0.348</td>
</tr>
</tbody>
</table>

When loaded 25 and 30kg, the carbon fiber pole showed faster recoil speed (84.09cm/sec and 82.69cm/sec) than glass fiber pole. Shown as Table 3.

Table 3 The difference between two materials' maximum dynamic pole recoil speed (unit: m/s)

<table>
<thead>
<tr>
<th>Load</th>
<th>Mean±SD</th>
<th>t</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>10kgw</td>
<td>10.80±6.00</td>
<td>1.80</td>
<td>0.102</td>
</tr>
<tr>
<td>15kgw</td>
<td>-3.21±5.98</td>
<td>-0.54</td>
<td>0.603</td>
</tr>
<tr>
<td>20kgw</td>
<td>1.05±3.35</td>
<td>0.31</td>
<td>0.761</td>
</tr>
<tr>
<td>25kgw</td>
<td>-2.30±3.85</td>
<td>-0.60</td>
<td>0.563</td>
</tr>
<tr>
<td>30kgw</td>
<td>-1.91±2.77</td>
<td>-0.69</td>
<td>0.506</td>
</tr>
</tbody>
</table>

The carbon fiber pole showed more stable performance on recoil speed. When loaded 10kg, the recoil speed ratio was 1.05, then became 0.90, 0.89, and 0.89 when weight increased to 20, 25 and 30kg; The speed ratio of glass fiber pole dropped from 1.00 to 0.89, 0.92, and 0.84 when load switched from 10kg all the way up to 20.25 and 30kg. Shown as Table 4.

Table 4 The difference between two materials' dynamic pole recoil speed (unit: m/s)

<table>
<thead>
<tr>
<th>Load</th>
<th>Mean±SD</th>
<th>t</th>
<th>p</th>
</tr>
</thead>
<tbody>
<tr>
<td>10kgw</td>
<td>-0.05±0.08</td>
<td>-0.61</td>
<td>0.556</td>
</tr>
<tr>
<td>15kgw</td>
<td>-0.09±0.05</td>
<td>-1.58</td>
<td>0.145</td>
</tr>
<tr>
<td>20kgw</td>
<td>-0.02±0.05</td>
<td>-0.36</td>
<td>0.724</td>
</tr>
<tr>
<td>25kgw</td>
<td>0.03±0.03</td>
<td>1.12</td>
<td>0.288</td>
</tr>
<tr>
<td>30kgw</td>
<td>-0.05±0.02</td>
<td>-2.22</td>
<td>0.051</td>
</tr>
</tbody>
</table>

Conclusions

The two materials were having similar amount of static and dynamic pole bend. The carbon fiber gave more resistance to vaulter who tried to bend it at the take-off phase. Once the carbon pole was bent, It would respond faster recoil speed, which would help vaulter to clear the bar with less effort.

References

The Comparison of High School and College Elite Swimmers for Start Performance

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Tunghai University, Taiwan²

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INTRODUCTION

Start is important for swimming performance. The grab start and track start are the most popular techniques over the world. There were many researches investigating the differences of the grab start and track start. The grab start was significantly higher than track start in impulse for collegiate swimmers, but track start was significantly higher than grab start in performance efficiency. In performance of 7m average velocity from the start, it did not showed the significant difference between two starts (S.T. Chen, W.T. Tang, H.J. Hsiao, 2006). However, the best performance for swimmers generally occurred in high school instead of university in Taiwan. Therefore, the purpose of this study was to compare the differences between high school and university athletes of swimming team in performance of grab start and track start.

METHODS

Twelve swimmers (6 university and 6 high school male swimmers) attend this study without injury. Each subjects performed three trials each starts individually. One digital video camera (JVC, 60Hz) and the force plate at 1200Hz (Kistler Instrument Corp., Winterthur, Switzerland) were synchronized and used to film the entry motion after start block and kinetic variables. Kwon3D 3.1 Motion System (VISOL Inc., Seoul, Korea) and Bioware 3.0 software (Kistler Instrument Corp., Winterthur, Switzerland) were used to analyze kinematics and kinetics data respectively. Efficiency is derived as the average velocity of 7m divided by the impulse. All kinetics variables were normalized by body weight. A 2 (age)×2 (start style) ANOVA was used to compare through SPSS 12.0 software.

Table 1  Information of two groups

<table>
<thead>
<tr>
<th></th>
<th>university (n=6)</th>
<th>high school (n=6)</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Height (cm)</td>
<td>179.42±7.64</td>
<td>174.25±4.38</td>
<td>0.181</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>72.93±4.71</td>
<td>67.17±6.26</td>
<td>0.101</td>
</tr>
<tr>
<td>Age (yrs.)</td>
<td>20.33±2.25</td>
<td>17.00±1.10</td>
<td>0.009</td>
</tr>
<tr>
<td>Train year (yrs.)</td>
<td>10.33±2.73</td>
<td>7.00±2.74</td>
<td>0.075</td>
</tr>
</tbody>
</table>

* : p<.05

RESULTS AND DISCUSSION

There is a significant difference between these two groups in age, but no significant differences either in height, weight and training years. (Table 1) And it may lead to the result that all of the parameters in this study did not show the significant difference between two groups. However, there is no more detail parameters (such as strength of lower limb) to find the real reason in this study.

For the aspect of two starts, there were significant differences between these two starts in impulse (1.20±0.12; 1.07±0.12) and efficiency (2.17±0.21; 2.41±0.34) but no significant differences either in 7m time, 7m average velocity, SSI and Fmax. The track start with higher performance efficiency compared to the grab start was due to the less effort (impulse) with the same level of performance(7m average velocity).

CONCLUSIONS

There were no significant differences between high school swimmers and university swimmers in this study. The techniques of start were equal for high school and university swimmers.

For the performance efficiency, the track start was significantly higher than grab start in efficiency. The results provide the reference to the coaches and swimmers the information for start training.

REFERENCES

THE RESPONSES OF CHILD USING ISOFIX CRS DURING IMPACTS


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Introduction

Accident data have demonstrated that a child restraint system (CRS) is effective for preventing injuries to children. Many studies have shown that the percentage of misuse is quite high in using CRS, and this misuse can limit the protective benefit of CRS. An ISOFIX CRS is effective for the reduction of the frequency of incorrect attachment of CRS to the car seat. Although child crash dummies are commonly used in vehicle crash tests, they have several limitations in biofidelity. The authors have developed a 3-year-old (3YO) child finite element (FE) model [1]. Using this model and Hybrid III FE model, a sled simulation of ISOFIX CRS was conducted and responses of both models were compared.

Materials and Methods

A 3YO child FE model has been developed by the authors for investigating injuries to children in impacts. Taking the anthropometry and material properties of a 3YO child into account, the model was made by scaling from the adult human FE model THUMS (Total Human Model for Safety). The responses of this child human FE model were validated in various impact conditions. In the child FE model, the skull shape was modified and the pelvis model was developed with Y-cartilage to represent child anatomy. A Hybrid III FE model provided by the First Technology Safety Systems was also used.

The sled tests according to the ECE R44 with 50 km/h velocity change were simulated. The behaviour was compared between child FE and the Hybrid III FE models. Injury criteria were also examined for the Hybrid III and child FE model. The responses of this child human FE model were validated in various impact conditions. In the child FE model, the skull shape was modified and the pelvis model was developed with Y-cartilage to represent child anatomy. A Hybrid III FE model provided by the First Technology Safety Systems was also used.

The kinematics and acceleration of the child in the ISOFIX CRS are shown in Figures 2 and 3, respectively. The ISOFIX attachment and the top tether secure the CRS effectively on the car seat, and make occupant restraint start early. The pelvis and shoulders are restrained by the lap and shoulder harnesses. Accordingly, the torso flexion angle and the head excursion were small. Due to head movement, the neck flexed and the chin made contact with the chest. As seen in Figure 3, whereas the thorax spine of the Hybrid III made of a steel box did not bend, the cervical and lumbar spine did.

Conclusions

Although there are differences in behaviour between the child FE model and Hybrid III FE model due to thorax spine flexibility in ISOFIX CRS, the injury criteria are similar for Hybrid III and child FE models. However, the child FE model can make it possible for detailed analysis of injuries to child.

References

EFFECT OF TARGET DISTANCE ON ANTICIPATORY POSTURAL CONTROL IN STANDING

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Introduction
Anticipatory postural control (APC) is defined as control of a sequence of postural adjustment reflected by muscle activity or center of pressure (COP) movement preceding a voluntary movement to minimize the potential postural perturbation caused by the forthcoming movement [1]. Results of previous studies showed that the APC is direction specific and affected by magnitude of perturbation [1, 2]. That is the dorsal side of muscle activated earlier than the ventral side in an arm forward raising movement. The onset latency of postural muscle is longer (earlier activation) in a maximal speed than in a self-paced speed; the COP shift backward earlier in a maximal speed than in a self-paced speed [1, 2]. However, previous studies which examined the effect of magnitude of perturbation were focused on the speed of arm raising movement. There is a limited study that examined such effect from the distance of reaching. The purpose of the study was to examine the effect of target distance on APC during standing.

Materials and Methods
2.1. Participants
Sixteen young adults (8 male, 8 female) participated in the study. They all reported to be healthy and had no deficit of balance.

2.2. Equipment
Postural stability was measured with a Kistler force platform (9286AA) and Bioware 3.2 data collection software. Six surface electromyogram (EMG) electrodes (MA317) were placed over muscles on the dominant side of the body (anterior deltoid (AD), biceps humerus (BH), tibialis anterior (TA), rectus femoris (RF), medial gastrocnemius (MG) and long head of biceps femoris (BF)) to measure the muscle activity. The equipment of force platform and EMG were synchronized and data were collected with a sampling rate of 1000 Hz.

2.3. Study design and procedure
Subjects were asked to perform a clinical test of functional reach test to determine their maximal forward reach distance [3]. Then they performed a reaching task in standing on Kistler force platform (Kistler a visual cue to press a button which was located at shoulder height level of four different distances: maximal forward reach (almf), 1/2 of maximal forward reach (alph), arm length (al), and arm length-1/2 of maximal forward reach (almh). Each task was repeated 5 trials and the sequence of task was randomized. The APC was expressed as the muscle activity 150 ms before and 50 ms after the activation of the primer mover (AD or BH). The center of pressure (COP) and vertical torque (Tz) were also measured and examined.

Results and Discussion
As predicted, the APC was related to the target distance. An early COP backward shift and vertical torque was associated with the distance of target (Figure 1, 2). The latency of postural muscle activity was also related to the distance of target (Figure 3).

Conclusion
Anticipatory postural control (an early COP backward shift and muscle activation) is associated with longer reaching distance.

References

Acknowledgement
The study was partially supported by a grant from China Medical University and National Science Council, Taiwan.
A STUDY OF MATERIAL AND FRICTIONAL PROPERTIES OF THE HUMAN FINGERTIP TO SIMULATE KINETIC FRICTION USING 3D FINITE ELEMENT MODEL

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** Digital Human Research Center, AIST, Tokyo, Japan
*** CREST, JST, Japan

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Introduction
Since the tactile perception detects skin deformation due to the contact of an object, understanding material and frictional properties of the human skin is essential to analyze the function of the fingertip quantitatively. In this study, we proposed a combination method using experimental and three dimensional (3D) Finite Element (FE) analysis to investigate material and frictional characteristics of the human fingertip.

Materials and Methods
Experiment using a tactile imaging system: A tactile imaging system was set up to capture an image sequence of the contact region between the fingertip and an indenter. We measured the following parameters: forces in the normal and tangential directions and contact areas of the index finger.

Development of 3D FE model: A 3D FE model of the index finger was developed after the image volume was segmented into skin, subcutaneous tissue, bone, and fingernail. The FE model has 8000 nodes and 7000 elements (Figure 1).

Material properties of the FE model: Young's moduli were 10GPa for the bone and 100MPa for the fingernail and both of Poisson's ratios were set at 0.3 (linear elastic for hard tissues) [1-2]. Each contact area of the FE model was optimized to correspond with that of the experiment using a Downhill Simplex method when the fingertip was indented into a flat plane with four kinds of depth, and we estimated material properties of soft tissues (both skin and subcutaneous tissue) which were presumed as hyperelastic.

Frictional property: When a surface has rubber-like material, the frictional behavior follows so-called rubber friction. A frictional equation \( \tau_{eq} = \lambda \sigma^\beta \) was hypothesized where \( \tau \) is the shear stress in the tangential direction (if the shear stress is more than the critical shear stress, slippage occurs), \( \sigma \) is the pressure, and \( \lambda \) and \( \beta \) are constants of friction. By using Downhill Simplex method for an optimization technique, two frictional constants were determined such that each "equivalent coefficient of friction" of the FE analysis corresponded with each coefficient of kinetic friction obtained from the experiment while the fingertip slid on a flat plane.

Results and Discussion
Table 1 shows the experimental results. Coefficient of kinetic friction was not constant and followed rubber friction.

Materials and Methods

Table 1 Results of the experiment

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>Normal force (N)</th>
<th>Coefficient of kinetic friction</th>
<th>Contact area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.75</td>
<td>0.120</td>
<td>2.46</td>
<td>49.4</td>
</tr>
<tr>
<td>1.00</td>
<td>0.198</td>
<td>2.00</td>
<td>62.8</td>
</tr>
<tr>
<td>1.25</td>
<td>0.384</td>
<td>1.53</td>
<td>83.2</td>
</tr>
<tr>
<td>1.50</td>
<td>0.454</td>
<td>1.42</td>
<td>97.7</td>
</tr>
</tbody>
</table>

The results to determine material properties of soft tissues are as follows. When initial shear moduli were 0.034MPa for the skin and 0.00684MPa for the subcutaneous tissue, the contact areas for the FE analysis were the closest to those for the experiment in Table 1. The estimated material properties agree with published data [1-2] and if individual variation is taken into account, the result of the FE analysis is appropriate.

We obtained the frictional equation as \( \tau_{\text{eff}} = 0.086 \sigma^{0.46} \) after optimization of frictional constants. Figure 2 shows the result of coefficient of friction. We simulated kinetic friction of the fingertip while sliding on a flat plane by using 3D FE analysis to estimate material properties of the soft tissues and to introduce the frictional law of the rubber-like material.

Conclusions
We simulated kinetic friction of the fingertip while sliding on a flat plane by using 3D FE model to estimate material properties and to introduce rubber friction. The FE analysis facilitates prediction of internal stress or strain and it allows us to understand the response of mechanoreceptors of tactile sensation while the fingertip is sliding.

References
Biomedical education systems for engineers: REDEEM and ESTEEM projects

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** New Industry Creation Hatchery Center, Tohoku University, Sendai, Japan  
matsuki@pfsi.mech.tohoku.ac.jp

Introduction

Current engineering applications in the medical arena are extremely progressive. However, it is rather difficult for medical doctors and engineers to discuss issues because they do not always understand one another's jargon or ways of thinking. To this end, Tohoku University in Japan has managed a number of unique re-education programs for working engineers. REDEEM (Recurrent Education for the Development of Engineering Enhanced Medicine) has been offered as a basic learning course since 2004, and ESTEEM (Education through Synergetic Training for Engineering Enhanced Medicine) as an advanced learning course since 2006.

The purpose of REDEEM is to train medical engineers to understand the ways in which medical doctors and scientists think. The curriculum has two goals. The first is to cover a broad range of topics from basic biology to clinical medicine, social medicine, and biomedical engineering. The second is to use three different learning systems (Lectures, Laboratory work, E-learning) to provide opportunities to systematically study biomedical engineering, from the basics to the higher levels.

ESTEEM is an advanced course. The objectives of this curriculum are to explain the ways in which medical doctors think and to create synergy between the business community and academia by breaking the barriers between them. The curriculum includes advanced clinical medicine, specifically surgery, for engineers. Lectures are ground-breaking in that medical doctors present and discuss their clinical cases in the format of a clinico-pathological conference, with students joining in the discussion as engineers. The laboratory work is also unique in that it consists of clinical diagnostic and therapeutic work. Students understand the meanings of various surgeries, how to make a diagnosis, and the use of surgical techniques as therapies. After the course, they are supposed to act as mentors at their companies for university interns. Experienced engineers with greater knowledge and technological expertise can create and direct new medical engineering businesses.

In this paper, we report on these programs and discuss their effects on working engineers.

Methods

REDEEM course In all, 41 students took the intensive REDEEM course in 2005–2006. Of the 41 students, 55% worked for private companies, and 24% for researchers at universities. With regard to age, 72% were in their 20s or 30s, and 19% were in their 40s. As for academic background, 45% had master’s degrees, 31% had bachelor’s degrees, and 24% had doctorates.

ESTEEM course In all, 20 students took the intensive ESTEEM course in 2007. Of the 20 students, 60% worked for private companies, and 25% for researchers in academia. With regard to age, 35% were in their 20s and 30s, and 45% were in their 40s. As for academic background, 30% had master’s degrees, 40% had bachelor’s degrees, and 25% had doctorates.

Examinations All students took a multiple-choice test before the first lecture to test their knowledge of biomedicine and clinical medicine; students answered the same questions after the final lecture to test the effect of the course.

Results and Discussion

Our programs attracted young engineers in particular, and those with master’s and doctoral degrees. The most important points of our programs were focused on the issues of whether students could obtain better outcomes and feel satisfied with the courses. Therefore, it was important to cater to engineers when developing the programs. In the end, we developed two unique, interactive biomedical educational systems (REDEEM, ESTEEM) for engineers, with five specific components as follows:

1. Modular Lecture 2. Disease-Based Lecture (Learning) 3. Case Study 4. Laboratory Work 5. Pre- and Post-tests

As for the effects of these courses, post-test score means were significantly higher than pre-test score means in every subject (Fig. 1a–e, Fig. 2a–b, p < .003, Wilcoxon’s signed rank test). Interactive lectures boost students’ motivation, interest, attention, and active participation and lead to increased satisfaction and better learning outcomes. And all students answered “satisfied” on a comprehensive evaluation of our programs. This objectively and subjectively shows that REDEEM and ESTEEM are effective biomedical education programs for working engineers. After taking these two courses, students could understand how medical doctors think about, and treat, humans; they also learned how they, as engineers, could use this knowledge and technology to develop medical instruments. They could also create and direct new medical engineering businesses. For this, continuing biomedical education is crucial.

![Fig. 1. REDEEM pre- and post-test scores](image)

![Fig. 2. ESTEEM pre- and post-test scores](image)

Conclusions

When developing a biomedical education program for engineers, it is very important to keep in mind engineers’ tendency to try to understand theoretical mechanisms through concrete examples. Tohoku University has developed two unique biomedical education programs for engineers, REDEEM and ESTEEM. As a result of taking these courses, all students obtained better outcomes objectively, on tests, and subjectively, through a measure of their satisfaction. Interactive, modular, and disease-based lectures or case studies are effective educational strategies, and laboratory work is crucial. Pre- and post-tests can be helpful not only in evaluating students’ achievements but also in enhancing the effects of lectures or learning.
Finite-Element Analysis of Pre-Burst Stress of Bursting Plant Pericarp

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** RIKEN, Computational Biomechanics Unit, Wako, Japan
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Introduction

The plant seeds are moved in a variety of ways, for example, carrying on the window, insect or animal. Dispersion by bursting pericarps is a typical moving way of plant seed that is self-active and independent of other creature (1). Pericarps, which can burst powerfully and spread the seeds widely, are advantageous to expand breeding grounds of the plant in natural selection process. So, pericarp bursting mechanism of extant plants seems to be optimized mechanically and structurally. Bursting motion of pericarp depends on its mechanical stress in self-equivalent condition just before bursting. In this study, bursting motion was tracked by high-speed video camera. Furthermore, pre-burst stress distribution of a pericarp was evaluted by finite-element method based on the pericarp shape before and after bursting.

Materials and Methods

We focused on burst fruit of Impatiens in this study. Fruit of Impatiens has five sheets of pericarp. The pericarps are connected each other and their boundary is stiffer than the body of pericarps. According to visual observation, bursting process of the pericarps is considered as follows. Pericarps start to absorb water and swell at the bursting season. Pericarps restrain their deformation each other, so swelling generate mechanical stress in pericarps. Once a pericarp boundary is ruptured by the stress, the pericarps deform quickly and throw the seeds outside. Burst motion of pericarp of Impatiens was taken by the high-speed video camcorder. And, we extracted track of three-dimensional motion of a pericarp during burst of the pericarp. Curves of a pericarp boundary were drawn based on images of high-speed video camera. CT scan of the pericarp was carried out by micro-CT. We constructed CAD models of the pericarp before and after burst based on the micro-CT images. We focused on a pericarp, which has ruptured boundary in the time of burst, and extracted surface data of the pericarp out of the CAD models. Then, Finite-element model was created for analysis by surface date of the CAD models.

We proposed reverse deformation procedure (2) to obtain stress distribution of the pericarp before burst. In the method, we assumed zero state stress in the pericarp after burst. If we give the pericarp reverse deformation starting from post-burst shape to pre-burst shape, stress condition of the pericarp just before burst must be attained. Add to this, we carried out computer simulation of dynamic burst motion of the pericarp by using finite-element analysis. Stress distribution obtained by reverse deformation analysis was given the pericarp model before burst as initial condition, in the simulation. Efficiency of the reverse deformation analysis can be demonstrated by comparing the simulated burst motion with actual one.

Results and Discussion

Figure 2(a) shows distribution of principal stress major in the pericarp at the time of just before burst. It corresponds to last increment of reverse deformation analysis. Compressive principal stress is higher in wide area of outside surface of the pericarp. On the other hand, tensile principal stress is higher in stem side half of inside surface. Low compressive stress region is observed in tip side half of pericarp inside surface. Figure 2(b) shows pericarp burst motion obtained by the simulation and actual burst motion of Impatiens fruit. Simulated burst motion seems to be corresponded well to the actual motion. Thus the pre-burst stress obtained by the reverse deformation analysis is probably effective.

Conclusions

Burst motion of Impatiens pericarp was taken by high-speed video camera, and boundary curves of a pericarp during burst process were extracted from the images. Micro-CT scan was carried out for the pericarps before and after burst, and finite-element models were created based on the CT images. Stress distribution of the pericarp before burst was obtained by the reverse deformation procedure using the finite-element models and deformation history of pericarp boundary. Furthermore, the pre-burst stress distribution was ensured by simulation of dynamic burst motion of the pericarp.

References