Evaluation of the mechanical characteristics of human thighs for developing complex dummy tissues

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***Abstract*— Wearable robots pose high risks to humans because they are directly mounted on human bodies, and methods for assessing their safety have not yet been established. For developing such safety tests, we have fabricated dummy legs that emulate the structure of the human body. Furthermore, all mechanical characteristics of human tissues that relate to their interactions with a wearable robot – elasticity, viscosity, normal stress, deformation, and friction coefficient – must be considered in parallel. Moreover, complex dummy tissues were developed and their characteristics were compared with those of human tissues. Mechanical characteristics of tissues for both humans and dummy were evaluated using a rheometer and an ultrasonic probe to clarify the similarities or differences. Consequently, we identified a mechanism of biological soft tissues as well as a way of imitating these characteristics in complex dummy tissues.**

***Index Terms*— Wearable robot, Viscoelasticity, Normal stress, Deformation, Friction coefficient, Dummy skin**

I. INTRODUCTION

A. Background

Recently, as aging of the population has rapidly accelerated, expectations for physical assistant robots to assist people with disabilities have increased [1][2]. However, issues such as safety standards, safety technology and cost reduction should be resolved to make assistant robots widely accepted to society. In particular, safety validation methodologies must be established as early as possible because these robots are directly mounted on the human body typically through cuffs, and there are risks of wounds or internal bleeding due to the interaction forces between the human and the robot [3]. The associated safety standards for personal care robots refer to the necessity of proper ergonomic design of physical assistant robots to ensure their operations without physical stress on discomfort [4]. In parallel with the safety standard formulations, some research groups have studied to clarify and measure human-robot interactions [5][6], but the successful development of safety criteria for wounds has not yet been achieved.

Our research group has focused on establishing a test method for estimating the wounds risk of wearable robots which are categorized in physical assistant robots. The risk is estimated as one of the highest due to the high frequency of occurrence in misalignment of wearing them. We develop a dummy leg that reproduces the characteristics and structure of the human body, and mount a wearable robot on the dummy to examine whether there is an ergonomic mismatch of kinematic motion between a human knee and a robot knee joint by quantitatively analyzing the interaction forces between them. For instance, we emulated the structure of human knee movement to achieve results similar to those for a human [7]. By using the dummy leg with urethane form covering, we can measure a similar tendency of interaction forces as those of a human. Nonetheless, we consider the dummy must be improved to estimate such wounds risks because the posture of the dummy leg is observed not to be necessarily same in the experiments due to a cause that the mechanical characteristics of the dummy skin are different from those of a human body. For improving the performance of the dummy leg, we currently concentrate ourselves on mimicking the mechanical characteristics of human tissue including its elasticity, viscosity, normal stress, deformation, and friction coefficient of human tissues.

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B. Research Subject

The mechanical characteristics of biological soft tissues have been extensively examined for a long time. In particular, the viscoelasticity of human tissue has long received attention [8]. Such elemental properties of human tissue as viscoelasticity depend on many factors; age difference [9], parts of a human body [10], skin disease [11] and so on. However, considering only the viscoelasticity to emulate the mechanical characteristics of human tissues in a dummy tissue is inadequate because many mechanical elements together play an important role for slipping or interaction forces between the human and a robot. In fact, no study has yet revealed the comprehensive mechanism of human tissue. We focus on all of the elemental properties of human tissue – elasticity, viscosity, normal stress, deformation and friction coefficient – that appear relevant to the mechanism of human tissues, for developing a more sophisticated dummy. In this study, we measured and evaluated the elemental properties of human tissues by using a rheometer and ultrasonic probe, and then developed dummy tissues that show characteristics similar to those of human tissues.

The paper is organized as follows. In section II, we describe an experimental method for quantitatively measuring the viscoelasticity and normal stress of human tissue. Then, we show the experimental results in section III. In section IV, for discussing the mechanical characteristics, we move on developing dummy tissues and comparing them with those of human tissue. We conclude the study in the final section.

TABLE I: Experimental conditions for viscoelasticity and normal stress

Frequency	0.1– 10 [Hz]
Displacement	10^{-2} [rad]
Axial force	5 [N]
Sampling points	Anterior, Lateral, and Posterior surfaces of thigh (16.6 cm from lateral epicondyle)
Parallel-Plate Geometry	20 mm in diameter and fixed with double sided tape
Displacement	15 (Anterior, Posterior), 10 (Lateral) [mm]
Sampling points	Anterior, Lateral, and Posterior surfaces of thigh (16.6 cm from lateral epicondyle)
Parallel-Plate Geometry	20 mm in diameter and fixed with double sided tape

Fig. 1: Experimental position

II. EXPERIMENTS

A. Viscoelasticity and Normal Stress

We conducted experiments in a healthy male subject aged 21[1]. The subject was asked to remain still during the measurement, and lying the upper body diagonally on a bed in supine position. For considering the influence of human muscles at the thigh, he was also asked to relax and strain the quadriceps muscle. All of the viscoelasticity values were measured on the tissue at human thigh by use of a discovery hybrid rheometer from TA Instruments® with a Peltier plate and parallel-plate geometry. We also measured the normal stress of the thigh when the displacement in the same direction was applied. Experimental conditions for the measurements of the viscoelasticity and normal stress on the thigh are shown in Table I; the experimental set-up and thigh position are shown in Fig. 1. The values of both viscoelasticity and normal stress were measured 15 times repeatedly at each of the sampling points.

B. Deformation of Human Thigh

An ultrasonic probe and compression tester are used to measure the deformation of skin, fat, and muscle by applying compressive force to the thigh. The sampling points are the anterior, lateral, and posterior surface of the thigh, which are placed at nearly the same height above the knee. For example, the lateral measurement point is located at 16.6 cm above the lateral epicondyle, which is in the middle of the area covered by the upper cuff of the wearable robot. The compressive force was varied from 0 to 10 N (in steps of 0.2 N from 0 to 1.0 N, 0.5 N from 1.0 to 3.0 N, and 1.0 N from 3.0 to 7.0 N, and 10 N).

[1]The experiments were conducted with the consent of the subject.

III. RESULTS

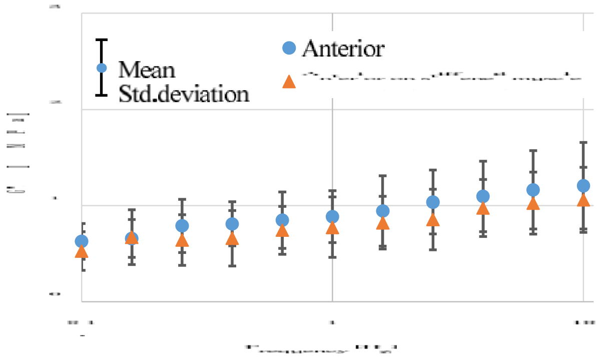
A. Viscoelasticity of Human Tissue

Elasticity and viscosity are given as the storage modulus $G'(\omega)$, and the loss modulus $G''(\omega)$, respectively. Fig. 2 shows that the experimental data are plotted in the format of mean $\pm$ standard error. The horizontal axis represents the frequency on logarithmic scale, which ranges from 0.1 to 10 Hz. We determined to focus on this frequency range because a numerical analysis of the interaction force exerted at the cuff in the frequency domain led us to a conclusion that the frequency components up to 10 Hz constituted a main lobe of 82.9 % in the histogram. The vertical axis indicates the elasticity and viscosity for Fig. 2a and Fig. 2b, respectively. For example, the mean values of the storage modulus on the anterior surface of the thigh are 0.88 $\pm$ 0.27 and 0.77 $\pm$ 0.31 MPa at 1 Hz, or 1.21 $\pm$ 0.45 and 1.06 $\pm$ 0.34 MPa at 10 Hz in the format of mean $\pm$ standard error, respectively. Successfully connecting the mean points in Fig. 2 reveals a slight increase of elasticity and viscosity with frequency at the thigh of the subject. Although the values of both the storage modulus and the loss modulus on the stiffened muscle are relatively lower than those on the relaxed muscle, we can conclude the differences were not significant for our objective of constructing a dummy tissue.

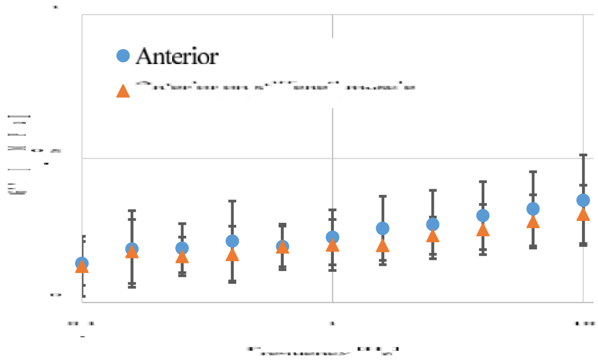
Data on the lateral and posterior surface of the thigh were also obtained to make comparisons. The elasticity and viscosity on the thigh parts are shown in Fig. 3 in the same format as in Fig. 2. Significant differences in the values of the storage modulus were observed in Fig. 3a. The storage modulus of the lateral of the thigh was relatively higher values than that of the posterior surface. In addition, the standard error of the storage modulus at the lateral surface is consistently high. In contrast, the loss modulus shows only a slight increase with frequency although the average loss modulus for the lateral surface is consistently higher than that for the posterior (Fig. 3b).

B. Normal Stress of Human Tissue

The normal stress against vertical displacement was also measured 15 times at the same three sampling points. In Fig. 4, the horizontal axis represents the displacement in the normal direction from the point where the parallel-plate geometry touches the skin surface. The vertical axis is the stress values obtained during the displacement. The normal stress for the lateral describes a more rapidly increasing curve than those for the anterior and posterior. The standard error of the lateral widens in proportion to the displacement increase. For comparing the values of the dummy tissue with those of

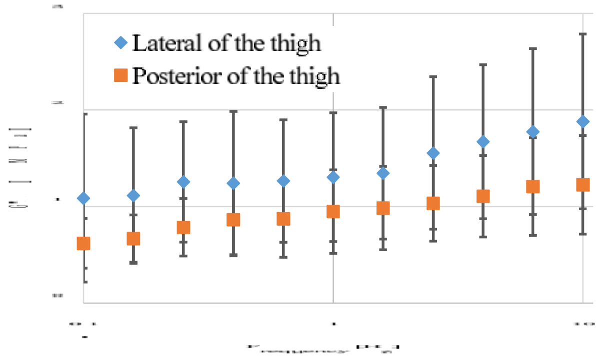


(a) Storage modulus

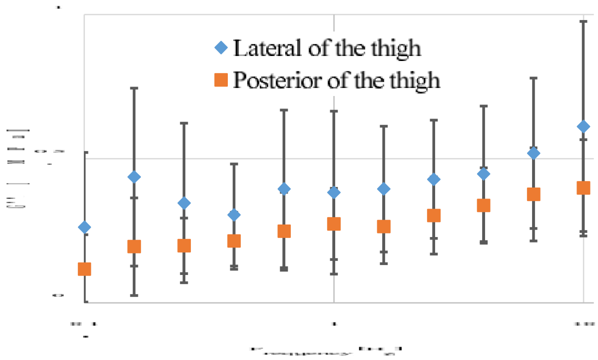


(b) Loss modulus

Fig. 2: Comparison of the storage modulus (a) or loss modulus (b) between a relaxed and strained quadriceps muscle



(a) Storage modulus



(b) Loss modulus

Fig. 3: Comparison of the storage modulus (a) or loss modulus (b) at the lateral and posterior of the thigh

human tissues, the focus in the section IV will be on the dummy tissue 1 and 2.

C. *Deformation*

For estimating factors (skin, fat, and muscle) that influence deformation of a thigh, images of partial sections were taken with an ultrasonic probe during the application of each compressive force. Fig. 5a is the thigh of a young male subject aged 22. The deformations of the anterior, posterior, and lateral of the thigh are shown in Fig. 5b, c, and d, respectively. Initially, the thicknesses of the skin, fat, and muscle are 37.3, 10.6, and 2.2 mm for the anterior; 44.1, 12.3, and 3.0 mm for the posterior; and 32.0, 4.7, and 2.4 mm for the lateral, respectively. At the normal force of 10N, the corresponding values are 24.2, 6.7, and 1.3 mm for the anterior; 24.2, 7.1, and 1.7 mm for the posterior; and 21.0, 3.3, and 1.6 mm for the lateral, respectively. The skin and fat undergo relatively little deformation, whereas the deformation of the muscle is observed in every part of the thigh.

IV. DISCUSSION

A. *Viscoelasticity*

The change of the stiffness of the muscles is not considered relevant to viscoelasticity of the surface as in Fig. 2 because subcutaneous fat presumably mitigates mechanical interference between the skin and muscle. In addition, the deformation of the muscle does not interfere with the shape and stress of the skin surface. Fig. 3a reveals a large standard error on the lateral of the thigh for the following reasons. First, the measurement method does not apply to the lateral surface because the thigh is not completely parallel to the parallel-plate geometry in the measurement for the lateral case. Second, an induration of the thigh apparently causes the large standard error because the values for the lateral surface show a small tendency of an induration to the vibratory motion of the probe. Further experiments are necessary to

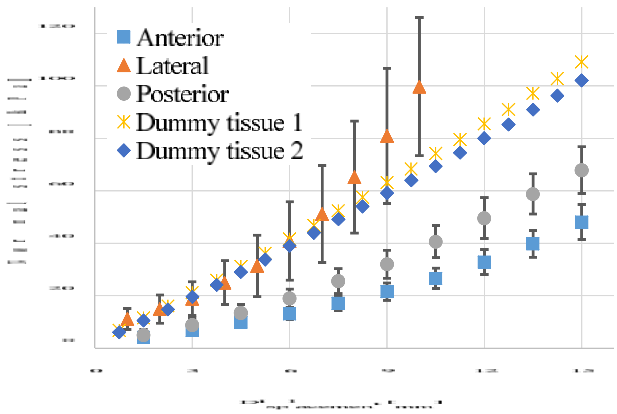
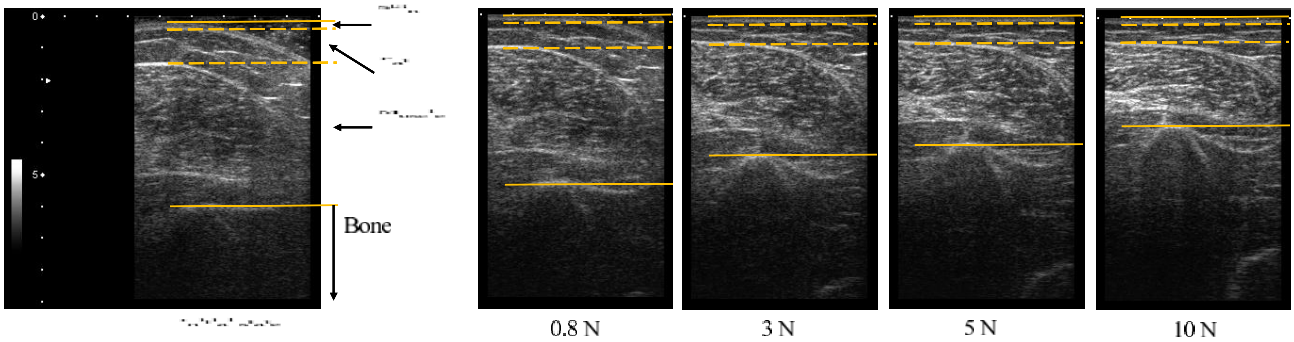
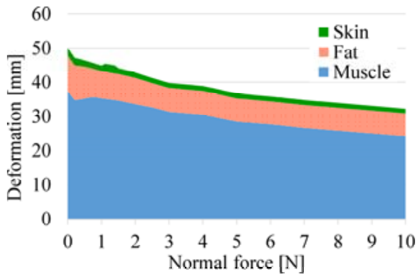


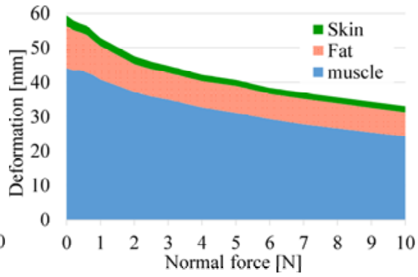
Fig. 4: Normal stress of the thigh to displacement



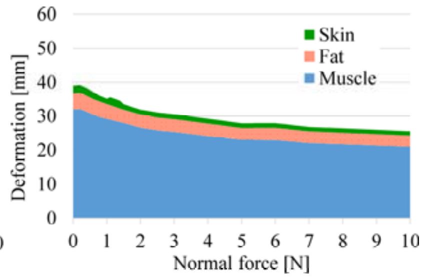
(a) The partial section



(b) Deformation of the anterior



(c) Deformation of the posterior



(d) Deformation of the lateral

Fig. 5: Deformations of (b) anterior, (c) posterior, and (d) lateral of the thigh

reduce the amount of the standard error by improving the measurement condition or finding more appropriate posture of the subject's lower extremity.

For emulating the viscoelasticity of human tissue, artificial soft tissues should be measured in the same way. We developed a complex dummy skin that consists of different materials : polyurethane gel, 5 mm thick with a hardness of 5 as measured by an ASKER Durometer Type C®, acrylic adhesive tape, and wound dressing. The complex dummy skin is structured so that the wound dressing is adhered with the acrylic on the polyurethane gel. We intentionally used three different materials because of the following three reasons. First, a polyurethane gel has shown a similar tendency of viscoelasticity with human tissue. Second, an acrylic adhesive tape can raise the values of viscoelasticity. Third, a wound dressing can mimic the friction of human skin.

Fig. 6 shows comparisons of the storage and loss moduli in human tissue and the complex dummy skin. The storage modulus of the dummy tissue was relatively lower than that of human tissue. We can improve the storage modulus by changing the adhesive force of the acrylic. However, both storage and loss moduli exhibit extremely similar tendencies even though each material shows different mechanical characteristics. We can conclude that the dummy skin emulate the storage and loss moduli of human tissue.

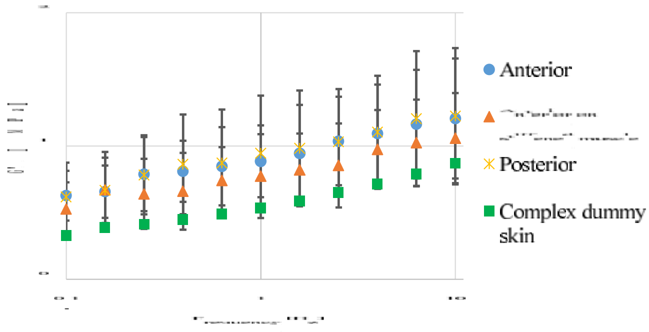
B. *Normal Stress and Deformation*

We developed complex dummy tissues 1 and 2, and their normal stress values are also shown in Fig. 4. Dummy tissue 1 was structured so that the complex dummy skin, which is mentioned in the previous subsection, was placed on a 45 mm polyurethane gel layers (hardness 5). In contrast, Dummy tissue 2 was structured so that the complex dummy skin was placed on a 55 mm polyurethane gel layers (hardness 5). The thickness of the dummy tissue was determined to facilitate comparisons between the dummy tissue data and data from the thigh because the thickness for the anterior and posterior was measured from 50 – 60 mm as shown in Figs. 5.

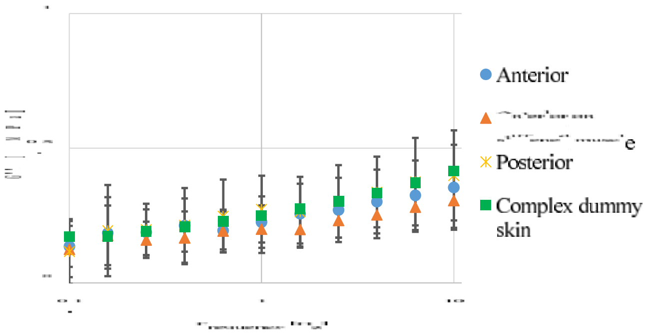
Fig. 5b, c, and d show that the skin and fat have little influence on the deformation of the thigh. Muscle plays an important role in thigh deformation, and human tissue displays a nonlinear tendency. However, the dummy tissues exhibit a proportional tendency to the displacement and hence cannot emulate human muscle in our current composition. Although the normal stress of the dummy tissues is between the values of the lateral and the posterior of the thigh in Fig. 4, a linear interpolation does not help for designing the thickness of the dummy tissues. Therefore, the hardness of the dummy must be further adjusted because the normal stress of the dummy tissues was higher than that of human thigh, or other materials for the human muscle part must be sought for higher fidelity of the nonlinear normal stress - displacement characteristics of human muscles.

C. *Friction coefficient on human skin*

Frictional property on human skin has also to be taken into consideration as a factor of biological characteristics of human tissues for estimating a less serious injury of wounds or internal bleeding. Naylor (1955) reported the first study on the human skin friction [12], and he showed that the coefficient of friction between the skin and polythene was



(a) Storage modulus



(b) Loss modulus

Fig. 6: Comparison of (a) storage modulus or (b) loss modulus for human tissues and dummy skin

0.5. Following this report, friction on human skin has been focused and studies from a medical attention or scientific interest [13][14]. While the interest for those frictional values has potentially been associated with friction blistering, regional differences, and its dependence on material and experimental conditions. Therefore, comparisons of the friction coefficient data in the literature are indispensable.

For estimating the friction coefficient of the dummy skin by using a rheometer, a tribological test was performed with the aid of reference [15]. We evaluated the static and dynamic friction coefficients of an annulus of the complex dummy skin with a parallel-plate steel geometry as shown in Fig. 7. The normal force F_N, constant angular velocity Ω, and gap H are measured by the encoder. The inner radius R_1 and outer radius R_2 are 15 and 20 mm, respectively.

Kavehpour [15] reported that the variation in shear rate across the annulus is small and the viscosity is constant in the case of $(R_2 - R_1)/R \ll 1$. Therefore, the average torque is given by

$$T = 2\pi \int_{R_1}^{R_2} \frac{\eta \Omega r^3}{H} dr = \frac{2\pi\eta\dot{\gamma}_{\overline{R}}}{\overline{R}} \left(\frac{{R_2}^4 - {R_1}^4}{4} \right)$$

where the average of the annulus is $\overline{R} = (R_1 + R_2)/2$, the average shear rate is $\dot{\gamma}_{\overline{R}} = \Omega\overline{R}/H$, and the viscosity is η. The friction coefficient is calculated as

$$\mu = \frac{T}{F_N \overline{R}}$$

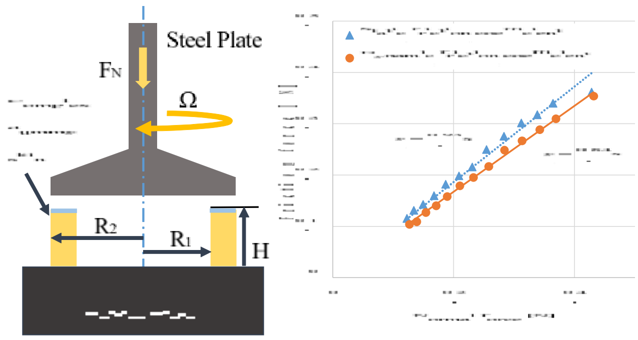


Fig. 7: Schematic diagram of testing method of friction. The static and dynamic friction coefficients are 0.93 and 0.84, respectively.

The average torque is exerted to increase in proportion to that of the strain and becomes maximal at which the static friction force is detected. Thereafter, the torque becomes constant at a lower value; at this point, the dynamic friction coefficient is observed. As shown in Fig. 7, the static and dynamic friction coefficients of dummy skin with viscoelastic material sheets under parallel-plate steel geometry, obtained by linear approximation, were 0.93 and 0.84, respectively.

We compare the static friction coefficient data with previously reported friction coefficient values for human skin. In [16], the static friction coefficient was shown to range from 0.60 to 0.89 at the skin-material interface, and from 0.48 to 0.66 at the skin-sock interface. EL-Shimi [17] showed that the friction coefficient was between 0.17 and 0.51 with a stainless steel probe, and noted that a friction force is not related linearly to a normal load, and does not obey Amonton's law. Friction coefficients of human skin that are measured in previous studies differ from each other and depend not only on experimental methods but also on environmental factors. Kwiatkowska [18] used a steel ball probe to measure the friction coefficient and showed relatively high values of 1.1 to 1.5 for the static friction coefficient and 0.7 to 1.2 for the dynamic friction coefficient. The differences are due to friction on the skin surface that arises from two mechanisms: interfacial adhesion and deformation [19]. On the other hand, a more resent study reported by Hendriks [20] discusses hydration dependence of human skin friction coefficients. In the literature, both static and dynamic friction coefficients are measured at human cheek parts by use of a rotating ring apparatus which is similar to the one in our study. The measured values are roughly 0.6–0.9 and 0.4–0.6, respectively at low hydration. Although the similarity between cheek and thigh has not been reported, the comparatively similar tribological test certifies the validity of our results. We can positively state that the friction coefficients of the complex dummy skin are detected appropriately.

V. CONCLUSIONS

We measured the mechanical characteristics of the anterior, lateral, and posterior surfaces of human thigh. We found that muscle stiffness had no notable influence on the viscoelasticity. The lateral of the thigh exhibited values higher than those of the anterior and the posterior, although the standard error of the lateral was large. Muscles played an important role in the deformation and normal stress of the thigh. We developed complex dummy tissues incorporating materials with different hardness and thickness with each other through a variety of preliminary experiments, and then compared their characteristics with those of human tissue in terms of viscoelasticity, normal stress, deformation, and friction coefficient. The complex dummy tissues have the mechanical characteristics similar to those of human skin. This complex dummy tissue can be used to supply our dummy leg with more sophisticated performance, which is expected to improve the quality of testing methods of wearable robot safety.

VI. ACKNOWLEDGMENT

The study was financially supported by NEDO (New Energy and Industry Technology Development Organization), Japan. The authors are indebted to Mr. Kenji Ishiguro, Research Associate Shogo Okamoto, and Associate Professor Susumu Hara, all with Department of Mechanical Science and Engineering, Graduate School of Engineering, Nagoya University, for their discussion on the study.

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