An estimation of the stiffness of mm. triceps surae in human hopping by the mass-spring model

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Abstract

Vertical component of ground reaction force (GRF) in rhythmical hopping motions of seven male subjects was recorded to analyze the stiffness of mm. triceps surae based on a simple mass-spring model. Hopping motions were performed only with plantar flexion, with knee and hip joints kept as straight as possible. Human body was modeled by a simple model, which was composed of a mass and a linear spring, which corresponded to human. Averaged difference between recorded GRF and theoretically predicted GRF based on the model ranged from 10 to 14% of the peak value of GRF, which was regarded as small. From the length of the phase in which GRF was larger than the body weight, value of the spring constant was estimated based on the simple model. The value ranged from $2.4 \times 10^4$ to $10.4 \times 10^4$ N/m, varying with the change of the hopping frequency. The reproducibility of the estimated stiffness, which was measured in different days, was quite high (CV ranged from 2.2 to 4.4%). Furthermore, predicted jump height, which was calculated from estimated spring constant, matched well to actually measured jump height. These facts supported the validity of the simple mass-spring system model.

Keywords: Mass-spring model, ground reaction force, stiffness of mm. triceps surae, muscle-tendon complex

Introduction

Viscoelastic properties in muscle-tendon complex (MTC) play an important role in enhancing both the effectiveness and the efficiency of human performance. Then, the role of elastic components has often been discussed in biomechanical studies of human movements. Some researchers stated that the elastic energy stored in the elastic component in counter movements was utilized to enhance the performance (Anderson and Pandy, 1993; Komi and Bosco, 1978; Voigt et al., 1995). Relevant basic information concerning mechanical properties of each component has been directly measured through many experimental studies, utilizing animals (Alexander and Bennet-Clark, 1977; Baratta and Solomonow, 1991; Caldwell, 1995; Ettema and Huijing, 1989; Ettema et al., 1990; Prilutsky et al., 1996; Proske and Morgan, 1987;...
Roeleveld et al., 1993; Sasaki and Odajima, 1996; Shadwick, 1990; Soest et al., 1995; Zuurbier and Huijing, 1992).

In the study of human motions, direct measurements from muscles or tendons is usually difficult from ethical points of view, and relatively few in vivo studies have been performed (Gregor and Abelew, 1994; Komi, 1990). In addition, as the mechanics of human motion is non-linear, a mere possession of elastic components (e.g. tendons) does not necessarily yield the elastic movement of macroscopic structures (e.g. motion of joints, body, etc.). Mechanical properties of macroscopic structures have to be investigated and quantified independently of properties of each component (Lafortune et al., 1996ab, Shibayama et al. 1998).

For these reasons, many studies of human motions have been based on model analysis of the movement of human body, at macroscopic levels, rather than dealing with each component. Muscle-tendon complex has been usually modeled as a connected system composed of a contractile, an elastic, and a viscous components. Cavagna (1970) and Shorten (1987) modeled human mm. triceps surae as the viscoelastic properties, and estimated the properties of each component, by analyzing the impulse response of the system. These studies were performed under “passive” conditions, in which subjects simply received initial impacts and the following responses of their body were recorded and analyzed. Subjects were instructed to maintain their equilibrium against external perturbations. The “passive” in this study was defined as the condition keeping muscle activity level.

However, in some kinds of free human motions, humans “actively” utilize the mechanical properties of the body to enhance mechanical and/or physiological performances. The active state of muscle determines the number of cross-bridge attachments in muscle. The number of cross-bridge attachments is related to the elasticity of muscle fiber in active condition. On the other hand, the length and the cross-sectional area of tendinous tissue usually determine the stiffness of its tissue. In order to investigate the role of elastic properties in natural human movements, properties of individual components had better calculated through the analysis of dynamic motions, in which subjects are intentionally utilizing elastic properties of the body to enhance the performances. Although Shibayama et al. (1998) discussed the human mm. triceps surae behavior in repetitive ankle bending exercises by using a forced oscillation model, there have been no studies on active and repetitive human hopping in the viewpoint of the stiffness.

Besides, jumping ability can be improved by skill training even if the architecture and the function of the muscle-tendon complex are in the same condition. The change of the stiffness of the total muscle-tendon complex, which can be determined by the timing and the muscle activity level of a contractile component, cause this improvement. So, it is important for athletes to evaluate the stiffness of total muscle-tendon complex including muscle activity level. Although Cavagna (1970) and Shorten (1987) estimated the elasticity of muscle-tendon complex in the different activity levels, these muscle conditions were absolutely passive that the subjects had to keep the activity level against to load. On the other hand, it can be said in the present study that the subject have to positively control the activity level through the commands of the central nervous system. The purpose of this study was to estimate the human mm. triceps surae stiffness during repetitive hopping motions in macroscopic scales by using a simple mass-spring model (see Appendix 1).

Methods

Subject: Seven male graduate students (body
mass 64.3 ± 5.8 kg, body height 1.68 ± 0.07 m, age 24.6 ± 2.4 years) participated in this study as subjects. All subjects submitted an informed consent form.

Task: Subjects performed rhythmic hopping motions, synchronized with the auditory signal of a metronome (SQ-77, Seiko, Japan), at 14 frequencies. 13 frequencies (from 120 jumps/min to 208 jumps/min) were synchronized with the metronome. One task was at the maximal voluntary frequency, in which subjects hopped as quickly as possible. Three trials were at the maximal task. Sequence of different frequencies was randomized. Hopping motions were only with plantar flexion of ankle joints, keeping knee and hip joints as straight as possible. Arms were crossed in front of the chest.

Data collection: These motions were on a force platform (9281B, Kistler, Switzerland) and vertical component of ground reaction force (GRF) was A/D converted (Lab-PC+, National Instruments Corporation, Japan) and stored into a personal computer (FM-V, Fujitsu, Japan) at 1000Hz. Data collection was started when the motion of subjects was synchronized with the auditory signal of the metronome, and at least 20 consecutive jumps were recorded. For one subject, the reproducibility was examined in each frequency of jumps on a different day.

Model: The stiffness of the total muscle-tendon complex change in the movement frequencies like at repetitive hopping, which was controlled by muscle activity level. By using the mass-spring model in the hopping, we can examine the stiffness of the spring stiffness in each frequency of movements. The vertical displacement of the mass center of body (MCB) fit well a sine wave. This simple model in the repetitive movement, in which the stiffness was mainly determined by the frequency, could be considered as more optimal than the forced oscillation model. Body of each subject was modeled as a connection of a rigid mass of m kg (body mass) and a light linear spring of a spring constant k_surae N/m (Figure 1). This value was regarded as the stiffness of mm. triceps surae (note that this value is not the stiffness of the Achilles tendon). When this system is released from the height (of MCB) at y = h; the point y = 0 was defined as the height where distal end of the spring touches the ground without any strain) without initial velocity in the field of gravity -g m/s^2, the spring contacts with the ground at the velocity

\[ \frac{dy}{dt_{contact}} = -\sqrt{2gh} \] (1)

After the contact, the motion of MCB is described by the equation

\[ m \frac{d^2y}{dt^2} = -k_surae y - mg \]

By solving this differential equation, with the initial condition (1), instantaneous position of MCB at the time t (s), during contact phase, can be described as

\[ y = -\frac{mg}{k_surae} + \Delta y \cos(\omega t - \Delta \phi) \]

where

\[ \Delta y = -\frac{mg + \sqrt{(mg)^2 + 2mgk_suraeh}}{k_surae} \]

\[ \omega = \sqrt{\frac{k_surae}{m}} \]

![Figure 1](Typical example of ground reaction force curve. Example of ground reaction force (GRF) curve was shown (subject D, at 184 jumps per minute).
\[ \Delta \rho = \arcsin \left( \frac{mg}{\sqrt{(mg)^2 + 2mgk_{\text{sur}}h}} \right) \]

where \( t = 0 \) represented the instant of the contact.

The elastic energy stored in the spring at the lowest point is

\[ E_e = \frac{1}{2} k_{\text{sur}} (\Delta y)^2 \]

And, from the law of energy conservation, jumping height can be represented as

\[ h = \frac{E_e}{mg} + \Delta y \quad (2) \]

All the equations and representations shown above can be derived theoretically.

Data reduction: The phase when GRF was larger than 3% of body weight was defined as the contact time, whereas the phase when GRF was smaller than 3% was defined as the flight time (Figure 1). The sum of the contact time and the flight time was defined as one hopping time. Jump height was calculated from the length of flight time.

\[ h = \frac{g}{8} T_{\text{flight}}^2 \]

where \( T_{\text{flight}} \) represents the flight time. The phase in which GRF was larger than the body weight was defined as half cycle. From the length of half cycle \( T_{\text{half}} \), stiffness of mm. triceps surae, \( k_{\text{sur}} \), was calculated as

\[ \omega = \frac{\pi}{T_{\text{half}}} \]

\[ k_{\text{sur}} = m\omega^2 \]

Predicted (founded on the mass - spring model) value of GRF was calculated, with this value and measured peak value of GRF (GRF\(_{\text{peak}}\)).

\[ \text{GRF}_{\text{theory}} = (\text{GRF}_{\text{peak}} - mg) \cdot \cos(\omega t') + mg \]

where \( t' = 0 \) represented the instant when GRF was its peak value. Recorded GRF was compared with this value. Mean deviation of these two curves was calculated as the following representation

\[ \text{Deviation} = \frac{1}{T_{\text{contact}}} \int (\text{GRF}_{\text{theory}} - \text{GRF}_{\text{recorded}})^2 dt / T_{\text{contact}} \quad (3) \]

where \( T_{\text{contact}} \) represented the contact time.

Displacement of MCB during contact time was calculated by integrating the GRF over the time, with the initial velocity described in equation (1). This value and estimated \( k_{\text{sur}} \) value were put into the equation (2), and predicted (according to mass-spring model) jump height was calculated.

These calculations were performed for ten consecutive hopping motions, for each frequency, each subject. Calculated value for ten individual jumps were averaged, and mean values were regarded as the representative parameters at that frequency, for each subject. For each subject, deviation between predicted and measured value of GRF for all frequencies was averaged, and that value was defined as the mean deviation for the subject. This value was calculated in order to evaluate the validity of the model. For each subject, theoretically predicted jumping heights for all the frequencies were compared with recorded ones, then the differences were statistically tested (paired t-test). Note that the recorded jump height was calculating flight time only, and predicted jump height was calculated using the GRF during contact time only. In this test, the slope value was not different from 1 (the slope of the identical line).

Comparison with previous studies: Values of \( k_{\text{sur}} \) estimated in this study was compared with the values presented in preceding studies (Cusack and Miller, 1979; Harley et al., 1977; Sasaki and Odajima, 1996; Shadwick, 1990; Voigt et al., 1995), which investigated animal and human tendon properties. In some of those preceding studies, stiffness of the tendon was presented in terms of Young's modulus (GPa). Those values were transformed into the stiffness of the "imaginary" Achilles ten-
don, with arbitrary cross sectional area 49mm² and length 0.364 m (Yamaguchi et al. 1990). 2 multiplied calculated values, in order to represent the values for two legs. After that, the values were multiplied by 1/9, to consider for the effect of the ratio of the length (Achilles tendon - lateral malleolus)/(metatarsal heads - lateral malleolus). The results were regarded as the stiffness of the “imaginary” mm. triceps surae. The largest and the smallest values of k_{surae} for each subject were selected, and mean values among subjects were calculated, then compared with preceding data.

Elastic bounce of the body was also analyzed through the same method as was presented in a preceding study (Cavagna, 1970). Precisely the same data collection procedure and the data analysis techniques were adopted. In this case, not the “active” hopping motion but the “passive” oscillation after landing (initial contact) was analyzed. Although Cavagna (1970) used three types of passive landing conditions from vertical jump (both legs, one leg, and one leg + load), the both legs landing was used in this study which was considered as relatively low activity level. The same subjects performed landing motions. Subjects jumped up and fell on the force platform without bending the knee and hip joints, with two legs. Calf muscles were kept in sustained contraction. Subjects relied slightly on external support to maintain the equilibrium after the initial contact. GRF was recorded at 100 Hz and the oscillation of lower limb was analyzed.

As the oscillation was the “damped oscillation”, the damping constant was estimated by solving the equation

$$\ln GRF_{peak} = \text{Const} - \frac{b}{2m} t$$

where b represents the damping coefficient (Ns/m) and t represents time (s). Then, the stiffness of the lower limb, k_{Cavagna}, was estimated by solving the equation

$$f = \frac{1}{2\pi} \sqrt{\frac{k_{Cavagna}}{m} - \left(\frac{b}{2m}\right)^2}$$

where f represents the frequency of the oscillation (Hz). Note that GRF_{peak} and f are the variables recorded with the force platform and b and k_{Cavagna} are the values calculated based on the model presented by Cavagna (1970). This procedure made it possible to estimate the stiffness of lower limb under passive condition. Each subject performed five landings, and five performances were averaged to calculate the mechanical properties for each subject. For all subjects, stiffness of the lower limb, k_{Cavagna}, was estimated, and the results were compared with the values originally calculated in this study, k_{surae}.

Results

There was a general tendency that as the frequency of hopping increased, flight time clearly decreased, whereas contact time was almost constant. Typical examples of the recorded GRF, one hopping time, half cycle, contact time, and flight time are shown in Figure 1. At the highest frequency, length of one hopping time, half cycle, contact time, and flight time were the shortest, and jump height was the lowest (Figure 2). The coefficients of variance of the estimated stiffness of each frequency in different day ranged from 2.2 to 4.4%. Also, the percent difference of the stiffness in each frequency, which was measured in different day, was 3.9 ± 5.3%. So, the high reproducibility was obtained in each frequency of jumps. The predicted curve was well fitted in the measured one (Figure 3). Also, the differences between peak values in two curves were also shown in Figure 3b. Averaged difference of the actually recorded GRF from the prediction of the model ranged from 10 to 14% of peak value (Table 1). It can be said that these differences were not so large.
As the hopping frequency increased, jumping height decreased (for example, from 0.103 to 0.010 m in subject D). At the same time, there was a tendency for the stiffness of mm. triceps surae to become larger as the frequency increased. Typical example of the relationship between jump height (frequency) and stiffness $k_{var}$ was plotted in Figure 4. We try to evaluate the stiffness (as dependent variable), related to muscle activity level (as independent variable, in detail, jump height and jump frequency). In Figure 4, the stiffness of the jumps controlled by metronome showed the strong correlation to jump height, except to the three jumps (see the circle in figure) by maximal voluntary frequency. The stiffnesses of these maximal jumps were positioned at slightly high level to the regression line of the jumps controlled by metronome. Then, it can be said that there was a different type of jumps between controlled and maximized. Range of estimated $k_{var}$ for all subjects was listed in Table 1. Typical example

![Figure 2](image-url)

Figure 2 Typical example of the relationship between jump height and one hopping, half cycle, contact time, flight time (Subj. D).

![Figure 3a](image-url)

Figure 3a Comparison of the recorded and the predicted GRF curves during hopping.

![Figure 3b](image-url)

Figure 3b Percent of difference of peak GRFs in the recorded and the predicted ones.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Deviation (%)</th>
<th>$k_{var}$ (N/m)</th>
<th>$k_{var}$ (N/m)</th>
<th>Slope</th>
<th>Difference</th>
<th>$k_{com}$ (N/m) (Preceding Method)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(Average)</td>
<td>(Lowest)</td>
<td>(Highest)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>A</td>
<td>13.4</td>
<td>$3.4 \times 10^4$</td>
<td>$7.0 \times 10^4$</td>
<td>1.00</td>
<td>N.D.</td>
<td>$3.5 \times 10^4$</td>
</tr>
<tr>
<td>B</td>
<td>12.9</td>
<td>$2.4 \times 10^4$</td>
<td>$8.1 \times 10^4$</td>
<td>1.15</td>
<td>N.D.</td>
<td>$3.4 \times 10^4$</td>
</tr>
<tr>
<td>C</td>
<td>14.0</td>
<td>$3.1 \times 10^4$</td>
<td>$7.5 \times 10^4$</td>
<td>1.05</td>
<td>N.D.</td>
<td>$3.1 \times 10^4$</td>
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<tr>
<td>D</td>
<td>12.0</td>
<td>$3.6 \times 10^4$</td>
<td>$6.1 \times 10^4$</td>
<td>0.98</td>
<td>N.D.</td>
<td>$3.3 \times 10^4$</td>
</tr>
<tr>
<td>E</td>
<td>10.0</td>
<td>$2.7 \times 10^4$</td>
<td>$8.1 \times 10^4$</td>
<td>0.77</td>
<td>N.D.</td>
<td>$3.0 \times 10^4$</td>
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<tr>
<td>F</td>
<td>10.5</td>
<td>$4.0 \times 10^4$</td>
<td>$10.4 \times 10^4$</td>
<td>0.88</td>
<td>N.D.</td>
<td>$3.1 \times 10^4$</td>
</tr>
<tr>
<td>G</td>
<td>12.6</td>
<td>$3.8 \times 10^4$</td>
<td>$7.2 \times 10^4$</td>
<td>0.83</td>
<td>p&lt;0.05</td>
<td>$3.5 \times 10^4$</td>
</tr>
<tr>
<td>Mean</td>
<td></td>
<td>$3.29 \times 10^4$</td>
<td>$7.34 \times 10^4$</td>
<td>0.952</td>
<td></td>
<td>$3.30 \times 10^4$</td>
</tr>
<tr>
<td>S.D.</td>
<td></td>
<td>$5.78 \times 10^4$</td>
<td>$7.54 \times 10^4$</td>
<td>0.132</td>
<td></td>
<td>$2.04 \times 10^4$</td>
</tr>
</tbody>
</table>
of the relationship between recorded and estimated jumping height was shown in Figure 5. When linearly regressed, the slope of regressed line was not significantly different from 1 (slope of the identical line), for 6 in 7 subjects (N.D.), as shown in Table 1.

Estimated values of the stiffness of mm. triceps surae, \( k_{\text{surae}} \) (mean value for the lowest and highest values, \( 2.4 \times 10^4 \) and \( 10.4 \times 10^4 \) N/m respectively), were generally similar to the values presented in preceding studies (Table 2).

Estimated value of \( k_{\text{surae}} \) based on the procedure presented by Cavagna (1970) (mean for seven subjects, \( 3.3 \times 10^4 \) N/m) are similar to the calculated lowest values of \( k_{\text{surae}} \). These values are also similar to the value presented by Cavagna in his original study (mean value, \( 3.7 \times 10^4 \) N/m).

**Discussion**

Body of subjects was simply modeled by a mass and a linear spring. Actual motion of subjects

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**Table 2** Comparison of mechanical properties of elastic components to the previous studies.

Values of stiffness presented in this study and preceding studies were compared. In this study, stiffness of the mm. triceps surae was estimated, and that value was compared with the value of preceding studies, which presented the Young's modulus of elastic components. Highest and lowest estimations of \( k_{\text{surae}} \) and \( k_{\text{surae}} \) calculated in this study were listed (mean value for seven subjects). Young's modulus was transformed into the stiffness of mm. triceps surae, utilizing the arbitrary value of cross-sectional area and length of Achilles tendon (Yamaguchi et al., 1990).

<table>
<thead>
<tr>
<th>Author</th>
<th>Year</th>
<th>Young's Modulus (GPa)</th>
<th>Stiffness (N/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Present Study</td>
<td>-</td>
<td>-</td>
<td>7.8 ( \times 10^4 ) human surae (highest)</td>
</tr>
<tr>
<td>Cavagna</td>
<td>1970</td>
<td>-</td>
<td>3.3 ( \times 10^4 ) human surae (lowest)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>-</td>
<td>3.3 ( \times 10^4 ) human surae (preceding method)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>-</td>
<td>3.7 ( \times 10^4 ) human surae (two legs)</td>
</tr>
<tr>
<td>Harley et al.</td>
<td>1977</td>
<td>9.0</td>
<td>6.3 ( \times 10^4 ) human surae (one leg + load)</td>
</tr>
<tr>
<td>Cusack and Miller</td>
<td>1979</td>
<td>5.1</td>
<td>2.7 ( \times 10^4 ) rat tail tendon</td>
</tr>
<tr>
<td>Shadwick et al.</td>
<td>1990</td>
<td>0.76</td>
<td>1.5 ( \times 10^4 ) rat tail tendon</td>
</tr>
<tr>
<td>Voigt et al.</td>
<td>1995</td>
<td>1.2</td>
<td>2.3 ( \times 10^4 ) pig digital extensor tendon</td>
</tr>
<tr>
<td>Sasaki and Odajima</td>
<td>1996</td>
<td>2.9</td>
<td>3.6 ( \times 10^4 ) pig digital flexor tendon</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>8.7 ( \times 10^4 ) bovine Achilles tendon</td>
</tr>
</tbody>
</table>
and the predicted motion of the model were compared in terms of GRF. Measured GRF was similar to the curve of theoretical prediction, although there were small deviations. In addition, stiffness of mm. triceps surae was estimated by analyzing the shape of GRF curve. With this value and vertical displacement of MCB during contact phase, jumping height was theoretically predicted. This prediction and the recorded jumping height, calculated from the flight time, were not regarded as statistically different for 6 in 7 subjects. The relatively large deviation of slope (ex. Subj. B, E, F and G) might be caused by an influence of the motion of knee and hip joints. It can be said, however, that values calculated from different two points of view (flight and contact phases) matched well. Then, the results, supported the validity of the model and data treatment employed in this study.

The elasticity of the muscle-tendon complex is mainly located in the tendinous tissues. Then, the stiffness in high activity level of this study compared to the values of the imaginary stiffness of mm. triceps surae calculated from the data presented in preceding animal tendon studies (Cusack and Miller, 1979; Harley et al., 1977; Sasaki and Odajima, 1996; Shadwick, 1990; Voigt et al., 1995) (Table 2). Although values presented in Table 2 varied slightly, this difference could be proper. Besides that, arbitrary values of the cross sectional area and the length of tendon, and anatomical conditions of foot segment, were utilized in the calculation of the imaginary stiffness. As well, the materials treated in preceding studies were not necessarily human's. These factors may have affected the estimation of the stiffness of imaginary mm. triceps surae and caused the differences. However, the results of this study were within the range of the preceding studies.

Individual anatomical (moment arm, shape of foot segment, etc.) and/or physiological (viscosity and individual stiffness of tendon, muscle, etc.) conditions were not included in this model. Calculated value represented the net effect of all structures, externally observed as the stiffness of mm. triceps surae. This general outline is similar to the one adopted by Lafortune et al. (1996ab) and Shibayama et al. (1998). Lafortune et al. discussed mechanical characteristics of the total body in shock absorption. Shibayama et al. discussed the behavior of muscle-tendon complex in m. gastrocnemius during repetitive ankle bending based on the forced oscillation model. In future studies, more anatomical and physiological conditions will have to be considered in the construction of the model.

In this study, values of $k_{sura}$ ranged from $2.4 \times 10^4$ to $10.4 \times 10^4$ N/m (Table 1). The muscle activity level changed in different frequencies of the hopping, even in one-contact phase at a constant frequency. The stiffness estimated in this study demonstrates the average value during one-contact phase. There was a tendency for $k_{sura}$ to increase as the frequency became higher (jumping height became lower), to reach the highest value at the frequency of the maximal-effort frequency. This tendency has been caused by the tension change of the contractile components (muscles) in different intensities of exercises. In low frequency hopping motions, muscle had to be largely lengthened and shorten itself in the contact phase, and this type of stretch-shortening lowered the value of the estimated stiffness, $k_{sura}$. On the other hand, in high frequencies, muscles had to be kept nearly isometric and this type of stretch-shortening made the value of $k_{sura}$ larger. Shibayama et al. (1998) explained the following results by using the model of an external forced mass-spring system: The lengths of the total muscle-tendon complex and the muscle itself in m. gastrocnemius became to be in the opposite phase with increasing frequencies during repetitive ankle bending. In other words, the muscle length was shorted in dorsiflexion
phase against increasing tension. The muscle was very active in this downward phase as was confirmed by the high EMG activities. In muscle-tendon complex, subjects can control the muscle activity level as the contractile components, whereas they can not greatly change the mechanical properties of elastic elements. It is natural to assume that the change of $k_{\text{sur}}$ observed in this study can be ascribed to the change of the muscle activity. In muscle tendon complex, contractile component is controlled in accordance with the property of elastic component, to realize the intended motions as a whole.

Human hopping was analyzed based on the mass-spring model in this study. However, if the muscle tendon complex of mm. triceps surae was a pure spring and did not provide mechanical work during exercise, continuous hopping motion would be impossible and jumping height would become lower and lower as the effect of viscous properties and friction. Then, muscles have to keep on providing mechanical work to the body in repetitive hopping motion as was explained by Shibayama et al. (1998). The model adopted in this study may have been simplistic in this point, because the vertical displacement of MCB fitted to the sine curve in this cyclic movement. However, this effect also may have distorted the estimation of the stiffness presented in this study.

Stiffness of lower limb calculated as described in the preceding study ($k_{\text{Cavagna}}$) was almost the same as the value presented in the original study (Cavagna, 1970). This fact certified the reliability of the calculation of $k_{\text{Cavagna}}$. Those values were similar to the stiffness of mm. triceps surae ($k_{\text{sur}}$) at low frequency, calculated in this study (Table 1). Under these conditions, “passive” and “active” human motions externally presented similar mechanical properties. On the other hand, when the frequency of the hopping was high, higher value of $k_{\text{sur}}$ was observed. It was suggested that at high frequency, the muscle activity was largely different from the value in the oscillation and the effect was observed as the difference of the stiffness. Measured stiffness can vary largely when the muscle activity level of lower limb is different. Especially, the quite high stiffness in the maximal jumps was observed in comparison with the jumps controlled by metronome as shown in Figure 4. It was considered that the high pre-activation from the central nerves system exited just before contact phase in maximal jumps. This result might be an evidence that human could intentionally control the muscle stiffness through the commands of the central nerves system. Referred to the Cavagna’s results (1970), the stiffness by one leg jump was larger than that by two legs jump. In other words, the high stiffness was observed even in Cavagna’s passive condition. However, from a neurophysiological viewpoint, the difference between passive and active conditions could be classified as the influences of the central nerves system and of the stretch reflex. Those conditions are usually affected by the following “task-dependent”. Even if the mechanical output is same, the sensitivity of the stretch reflex and the activate level of central nerves system are changed by passive and/or active conditions. Then, it can be said that those conditions affect the tonus and the stiffness of muscle. Also, since it is difficult for subject to maintain the posture in one leg jump, the reproducibility of the jump shown by Cavagna might be not so high. Furthermore, there may be different conditions between the keeping muscle activity level and the controlling positively the level against to jumping frequency by the central nervous system. It is quite easy for subject to control the muscle activity level of this method rather than Cavagna’s one. There is a possibility that it is necessary to select the optimal method of the assessment of mechanical properties of human body, in accordance with the scope of the study. For example, if it is intend-
ed to assess the elasticity of lower limb during highly dynamic motions such as sports and exercises, it may be inappropriate to measure the properties under static conditions. In that case, it would be necessary to assess the properties under "active" conditions, as presented in this study.

Appendix 1.

There were the following backgrounds regarding to the elasticity of MTC in the present study. According to Hill's classical model, MTC consists of a contractile, a series and a parallel elastic components. The stiffness is defined as the ratio of delta force / delta lengthening of MTC.

1. The stiffness of the contractile component is related to the number of the cross-bridge attachments in muscle fiber, and is almost correlated to activity level of muscle.

2. The stiffness of the contractile component in fully activated muscle shows 5 - 10 times of that of tendon in the MTC.

3. If two different linear springs (the spring constant: K1 and K2, respectively) connect in series, the total stiffness constant is shown as K1 x K2/(K1 + K2). The stiffness of this model (total MTC) is mainly influenced by activity level in low activated muscle. Also, the stiffness in high-activated muscle closes to the tendon stiffness (Shorten 1987).

4. It was usually hypothesized in simple vibration model that the spring stiffness of mm. triceps surae is a constant. Based on this hypothesis, Cavagna (1970) have determined the MTC stiffness in 'passive' condition of muscle during the landing of vertical jump.

5. On the other hand, it was considered in the 'active' jumping of this study that the stiffness of MTC mainly based on the number of the cross-bridge attachments must be changed even in one take-off.

6. Based on these backgrounds on the stiffness, we try to determine the average stiffness during take-off of jumping by using the simple vibration model through the hypothesis that spring stiffness was not so change during the take-off phase.

References


マスースプリングモデルによるホッピング中の下腿三頭筋の硬度推定

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要約

連続その場跳躍の垂直床反力を男性7名の被検者について測定し、マスースプリングモデルを基に分析した。跳躍動作は、股・膝関節をできるだけ伸展し、足底屈のみで行うものであった。身体を質点と下腿三頭筋による線形スプリングからなる単純なモデルと仮定した。床反力の実測値とモデルを基にした推定値の差は、床反力の最大値の10-14%の範囲にあり、小さいものであった。床反力が体重レベルを越える局面の時間から、モデルを基に体のパネ定数が推定された。その値は、2.4 x 10^7 - 10.4 x 10^7 N/mの範囲にあり、連続跳躍のピッチに伴い変化した。パネ定数から計算した跳躍高は、7名中6名が実測値に近いものであった。これより、マスースプリングモデルの妥当性を示すものである。

キーワード：マスースプリングモデル、床反力、下腿三頭筋の硬度、筋脈複合体

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