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Manipulability Analysis of Lower Extremities Based on Human Joint-Torque Characteristics

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1. INTRODUCTION

A human performs a variety of skillful movements by adjusting dynamic characteristics of his/her musculoskeletal system according to target tasks and environmental conditions. In braking an automobile, for instance, a driver controls the brake pedal with his lower limbs by regulating foot force as well as leg posture adequately so as to slow down comfortably. Hence, there is a close relationship between posture and joint-torque in human multi-joint movements.

In the field of robotics, there have been several methods to evaluate the manipulability of robots with a multi-joint mechanism from the kinematical and dynamical viewpoints [1]–[3]. These methods can provide quantitative analysis of the robot based on its link configurations, and have been applied to the analysis of human movements [4] [5] and the manipulability evaluation of welfare equipments [6]. However, these previous methods cannot consider kinetic characteristics of the human musculoskeletal system that should be taken into account for investigating human movements.

On the other hand, there have been some researches on the relationship between the joint torque and angle in uniarticular movements of the lower extremities. For example, Yamasaki et al.[7] measured the maximum voluntary joint-torque in isometric condition, and examined the differences according to age, gender, and other physical characteristics. Miura and Hisamoto et al. [8] investigated the maximum joint-torque of the lower limbs with respect to the joint angle, focusing on the elderly. However, there is no evaluation method for the quantitative analysis of human multi-joint movements in consideration of human joint-torque characteristics.

The present paper proposes a new performance index inspired by human joint-torque characteristics based on experimental results with human subjects, and develops a method to quantitatively evaluate human multi-joint movements. This paper is organized as follows: Section 2 explains the proposed method based on human joint-torque characteristics. Section 3 investigates human joint-torque characteristics by a set of experiments with human subjects, and demonstrates the effectiveness of the proposed methodology through analyzing human leg movements.

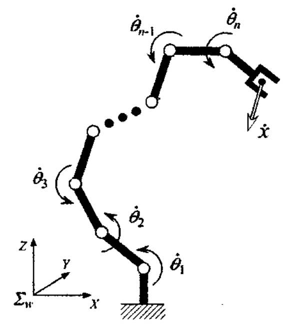


Fig. 1. An n-joint robot manipulator

2. MANIPULABILITY WITH HUMAN JOINT-TORQUE CHARACTERISTICS

In the field of robotics, there have been several methods to evaluate the manipulating ability of multi-joint robotic mechanisms [1]–[3]. These previous methods, however, cannot consider any biological characteristics of human motor movements. This section proposes a new evaluation method inspired by human joint-torque characteristics.

2.1. Force manipulability ellipsoid [1]

Figure 1 shows a general multi-joint robotic manipulator with n rotational joints in the m dimensional task space, where the joints are numbered from the basis in order, $\theta_1, \theta_2, ..., \theta_n$, and the arm posture can be represented with a vector $\boldsymbol{\theta} = (\theta_1, \theta_2, ..., \theta_n)^T \in \Re^n$. The relationship between a force exerted on the tip $\boldsymbol{f} \in \Re^m$ and a joint-torque $\boldsymbol{\tau} \in \Re^n$ is given by

$$\boldsymbol{\tau} = J(\boldsymbol{\theta})^T \boldsymbol{f}, \tag{1}$$

where $J(\boldsymbol{\theta}) \in \Re^{m \times n}$ is the Jacobian matrix on the end-point position $\boldsymbol{x} \in \Re^m$ with respect to $\boldsymbol{\theta} \in \Re^n$.

A set of the force $\boldsymbol{f}$ under $||\boldsymbol{\tau}|| \leq 1$ can be represented as an m dimensional ellipsoid given by

$$\boldsymbol{f}^T J J^T \boldsymbol{f} \leq 1. \tag{2}$$

The manifold on $\boldsymbol{f}$ in (2) is so-called force manipulability ellipsoid (FME) [2]. This has been used as a quantitative measure for the evaluation of manipulating ability on multi-body robotic mechanisms. The shape of this ellipsoid indicates a performance index in generating force according to the operational direction under the link posture $\boldsymbol{\theta}$. For example, large operational force can be easily exerted in the major axis direction, while it is difficult toward the minor axis direction.

2.2. *Biologically-inspired force manipulability ellipsoid*

Let us introduce the joint-torque activation level $\boldsymbol{\alpha} = (\alpha_1, \alpha_2, \cdots, \alpha_n)^T \in \Re^n$, in which $|\alpha_i| \leq 1$ represents the ratio of i-th joint-torque to the maximum torque under the maximum voluntary contraction (MVC) and its sign denotes the joint rotational direction (the flexional direction is defined as positive). In this paper, the joint-torque vector $\boldsymbol{\tau}$ is then expressed with $\boldsymbol{\alpha}$ because the muscle tension is proportional to its muscle activation level as follows:

$$\boldsymbol{\tau} = \boldsymbol{T}(\boldsymbol{\theta})\boldsymbol{\alpha} \tag{3}$$

where $\boldsymbol{T}(\boldsymbol{\theta}) = \text{diag.}(\tau_{1j}^{max}(\theta_1), \tau_{2j}^{max}(\theta_2), \cdots, \tau_{nj}^{max}(\theta_n))^T \in \Re^{n\times n}$ ($j \in \{f, e\}$; the suffix f and e indicate the flexional direction and the extensional direction, respectively). Each diagonal element of $\boldsymbol{T}(\boldsymbol{\theta})$ indicates an absolute value of the maximum joint-torque at the angle θ_i.

Substituting (1) into (3), $\boldsymbol{\alpha}$ can be represented with $\boldsymbol{f}$ as follows:

$$\boldsymbol{\alpha} = \boldsymbol{T}^{-1}\boldsymbol{J}^T\boldsymbol{f}. \tag{4}$$

A set of the force $\boldsymbol{f}$ generated by the muscles within $||\boldsymbol{\alpha}|| \leq 1$ then makes an m dimensional ellipsoid represented by

$$\boldsymbol{f}^T(\boldsymbol{J}\boldsymbol{T}^{-1})(\boldsymbol{J}\boldsymbol{T}^{-1})^T\boldsymbol{f} \leq 1. \tag{5}$$

Comparing the equation (2) and (5), it can be seen that the proposed ellipsoid is corresponding to the transformed FME by the matrix $\boldsymbol{T}(\boldsymbol{\theta})$ that reflects the characteristics of human joint-torque .

The principal axes of the ellipsoid can be obtained by a singular value decomposition of $(\boldsymbol{J}\boldsymbol{T}^{-1})$ as

$$\boldsymbol{J}\boldsymbol{T}^{-1} = \boldsymbol{U}_b\boldsymbol{\Sigma}_b\boldsymbol{V}_b^T, \tag{6}$$

where $\boldsymbol{U}_b = [\boldsymbol{u}_{b1}, \boldsymbol{u}_{b2}, \cdots, \boldsymbol{u}_{bm}] \in \Re^{m\times m}$, $\boldsymbol{V}_b \in \Re^{n\times n}$ is orthogonal matrices, $\boldsymbol{\Sigma}_b = [\,\boldsymbol{\Sigma}_{b0} \mid \boldsymbol{0}\,] \in \Re^{m\times n}$, $\boldsymbol{\Sigma}_{b0} = \text{diag.}(\sigma_{b1}, \sigma_{b2}, \cdots, \sigma_{bm}) \in \Re^{m\times m}$ ($\sigma_{b1} \geq \sigma_{b2} \geq \cdots \geq \sigma_{bm} \geq 0$). The principal axes can be then derived as $\boldsymbol{u}_{b1}/\sigma_{b1}, \boldsymbol{u}_{b2}/\sigma_{b2}, \cdots, \boldsymbol{u}_{bm}/\sigma_{bm}$. In addition, the volume of the ellipsoid V_{bf} can be obtained by

$$V_{bf} = c_m/w_b \tag{7}$$

where

$$w_b = \sqrt{\det[(\boldsymbol{J}\boldsymbol{T}^{-1})(\boldsymbol{J}\boldsymbol{T}^{-1})^T]} = \sigma_{b1}\sigma_{b2}\cdots\sigma_{bm}.$$

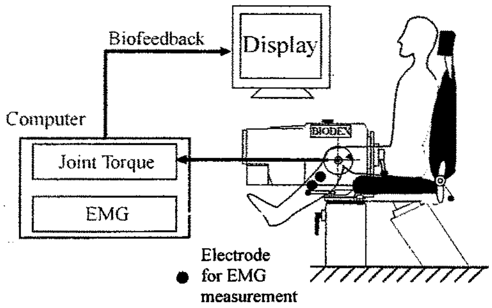
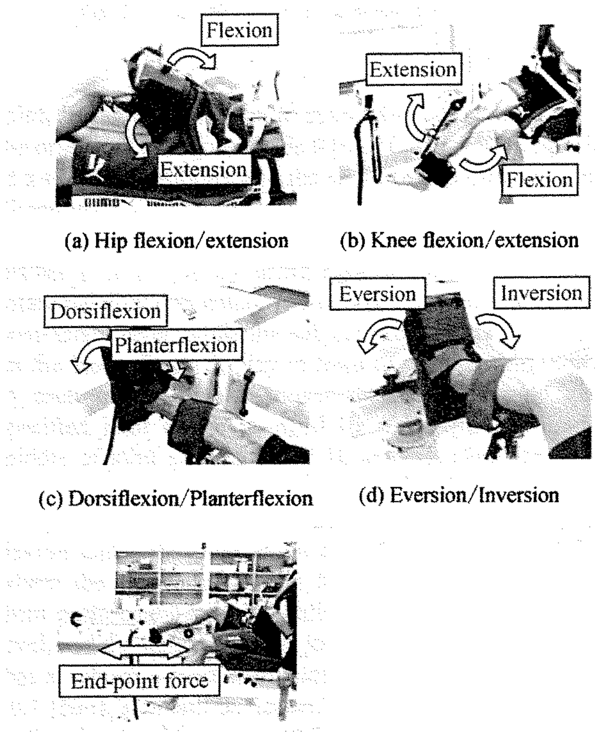


Fig. 2. Experimental apparatus



(a) Hip flexion/extension (b) Knee flexion/extension

(c) Dorsiflexion/Planterflexion (d) Eversion/Inversion

(e) Closed chain

Fig. 3. Mesurement of joint-torque and end-point torque

3. Manipulability Analysis of Lower Extremities

In order to apply the proposed method to the analysis of human movements, it needs to provide the maximum joint-torque of each uniarticular movement, $\tau_{ij}^{max}(\theta_i)$, which is a diagonal element of the matrix $\boldsymbol{T}(\boldsymbol{\theta})$. This section first investigates human joint-torque characteristics with human subjects to determine τ_{ij}^{max}, and then demonstrates the analysis of human leg movements to show the validity of the proposed method based on the experimental results.

3.1. *Joint torque measurement experiment*

1) Experimental apparatus: Figure 2 illustrates the experimental apparatus for investigating human joint-torque characteristics, which consists of the measurement part of joint-torque and EMG signals, and the bio-feedback display

TABLE I
AGONIST FOR EACH OF UNIARTICULAR MOVEMENTS

Uniarticular movement	Agnoist
Hip flexion	iliacus
Hip extension	musculus gluteus maximus
Knee flexion	musculus semitendinosus *
Knee extension	musculus rectus femoris *
Dorsiflexion	gastrocnemius *
Planterflexion	anterior tibial muscle *
Evertion	anterior tibial muscle *
Inversion	long fibular muscle

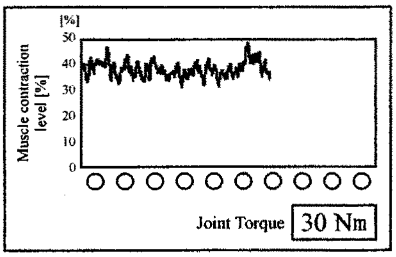


Fig. 4. Biofeedback display

in order to monitor the muscle activation level during experiments. In this paper, BIODEX-SYSTEM-2AP (MEDICAL SYSTEMS INC.; the maximum torque is 610 [Nm]; the maximum angular velocity is 7.854 [rad/s]) was employed for measuring joint-torque. A set of attachments was assembled according to target movements as shown in Fig. 3, where the figure (a) is for the hip flexion and extension; the figure (b) for the knee flexion and extension; the figure (c) for the dorsiflexion and planterflexion; the figure (d) for the eversion and inversion; and the figure (f) for the end-point force. The EMG signals were measured from the agonists as shown in Table I with an amplifier (NEC Medical Systems, MT11) and a set of disposable electrodes (GE Marquette Medical Systems Japan, Biorode SDC-H). The agonists with the asterisk in Table I were selected in a case of measuring the end-point force by the lower limbs.

The measured EMG signals were rectified and integrated with the data during the past 0.1 seconds, and then normalized with the values measured in the MVC at the middle angle of the range of joint motion. The normalized value was utilized as the muscle contraction level in this paper.

Figure 4 illustrates an example of the biofeedback display, where the joint-torque and the muscle contraction level are presented while the circles are filled from the left every second to indicate the measured time to a subject.

2) Experimental method: The lower extremities were modeled with a rigid link model with four rotational joints as shown in Fig. 5. In the figure, $\theta_1, ..., \theta_4$ represent joint angles of the hip flexion and extension, the knee flexion and extension, the dorsiflexion and planterflexion, and the eversion and inversion, respectively. l_1, l_2, l_3 denote link length from hip joint to knee joint, from knee joint to ankle joint, and from ankle joint to ankle. The dotted lines indicate the origins of each joint angle 0 [rad] (an anatomical position at a standing position), and the arrows are positive rotational directions.

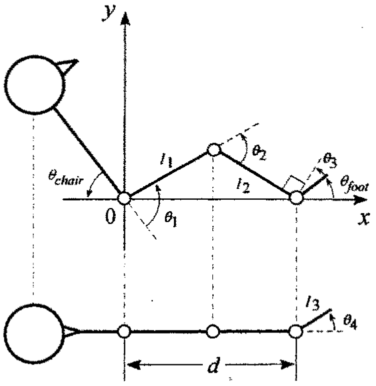


Fig. 5. The link model of the human lower extremity

In the experiments, a subject is instructed to generate his/her joint-torque by using agonist toward the specified rotational direction without co-contraction at the 40% muscle contraction level, while the subject can monitor the progress on the biofeedback display in front of him/her. Joint torques in each of uniarticular movements are measured at the specified joint angle, changed by 0.349 [rad] around the middle of joint excursion, for 10 seconds (See Fig. 3(a) ~ (d)). Two trials were conducted at each joint angle.

Figure 6 shows an example of the measured data in plantar flexion under the joint angle of ankle was at 0.349 [rad], where the top figure is the time history of EMG signal from gastrocnemius, the middle one the muscle contraction level, and the bottom one the joint-torque. It can be observed that a subject almost maintains the joint-torque (average: -30.1 [Nm], standard deviation: 0.882 [Nm]) and the muscle contraction level (average: 36.8 [%], standard deviation: 5.27 [%]) for ten seconds.

The measured data was processed along the following procedure:

i) Divide the signals measured during 10 seconds by every second in each trial, and calculate the average and standard deviation of the muscle contraction level.
ii) Extract 10 periods in which the muscle contraction level is close to 40% from the 20 periods created in the previous process.
iii) Calibrate the average joint-torque of the extracted sections to one at the 40% muscle contraction level.
iv) Calculate the average and standard deviation of the calibrated joint torques.

The measurement experiments were carried out with the four male subjects under the conditions mentioned above.

3) Experimental results: Figure 7 represents a typical example of the measured joint-torque of Subject A in hip flexion and extension within 0.087 ~ 2.182 [rad], the knee flexion and extension within -2.182 ~ -0.087 [rad], the dorsiflexion and planterflexion within -0.698 ~ 0.349

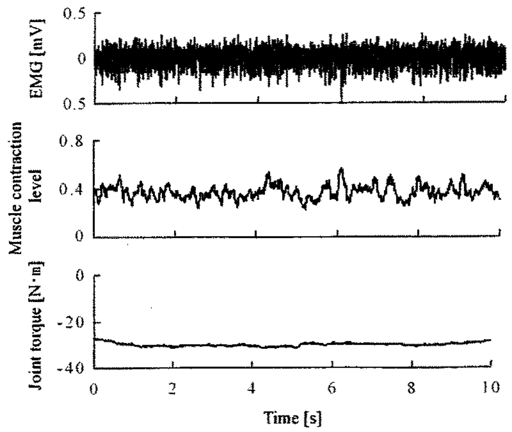


Fig. 6. Example of measured signals

[rad], and the eversion and inversion within -0.698 $\sim$ 0.698 [rad]. A solid line with white circles represents the joint-torque in the hip and knee flexion, dorsiflexion, eversion, while a dotted line with black circles in the hip and knee extension, planterflexion, and inversion. Almost the same characteristics were observed on other subjects.

Through observations of the experimental results, two tendencies of the joint-torque characteristics have been found in the uniarticular movements: (i) the joint-torque is almost proportional to the joint angle as in the hip flexion and extension (Fig. 7 (a)), the knee flexion (Fig. 7 (b)), and the inversion (Fig. 7 (d)). It can be supposed from the biological evidence that the muscle can generate larger force as the length of muscle longer because of its stiffness effect. (ii) the joint-torque peaks at the neutral angle of range of joint motion as in the dorsiflexion and planterflexion (Fig. 7 (c)). It can be supposed that it is difficult to generate large torque around boundary of the joint excursion.

3.2. *Estimation of the maximum joint-torque* $T(\theta)$

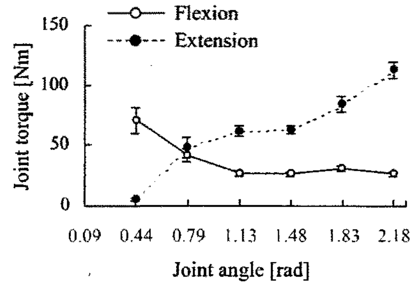
The maximum joint-torque in each uniarticular movement under the MVC, $\tau_{ij}^{max}(\theta_i)$, was calculated from the measured data at 40 % muscle contraction level by magnifying two and a half times under the assumption that joint-torque is almost linear to the muscle activation level of the corresponding agonist in this paper. Then, $\tau_{ij}^{max}(\theta_i)$ was modeled with respect to joint-angle θ within the measured range of joint motion by a third order polynomial as follows:

$$\tau_{ij}^{max}(\theta_i) = a_3\theta^3 + a_2\theta^2 + a_1\theta + a_0, \tag{8}$$

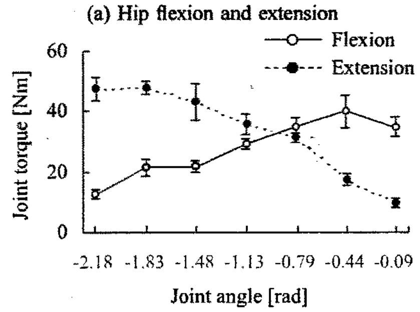
where the coefficients $a_0 \sim a_3$ were estimated with the least squares method.

3.3. *Force manipulability in lower extremities*

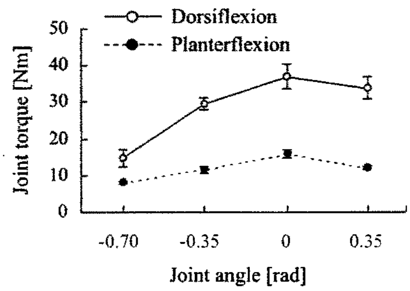
The validity of the proposed method was demonstrated though evaluating the force manipulability of human leg movements. First, the characteristics of end-point force generated by the leg limbs were shown from experiments with the subjects.



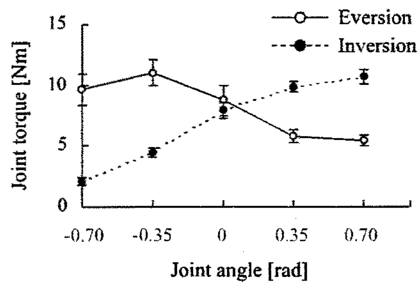
(a) Hip flexion and extension



(b) Knee flexion and extension



(c) Dorsiflexion and planterflexion



(d) Evertion and inversion

Fig. 7. Joint-torque characteristics with respects to joint-angle and rotational direction in the lower limbs (Subject A)

1) End-point force by Human leg movements: In the experiments, a subject was instructed to generate end-point force toward the x direction by using agonist without co-contraction under 40 % muscle contraction level (See Fig. 3(e)). The end-point force was measured twice at the distance d changed by 0.05 [m] (See Fig. 5), where the positions of trochanter major and malleolus were set at the same height under θ_{foot} = 0.873 [rad] and θ_{chair} = 0.916 [rad].

Figure 8 shows an example of the end-point force by Subjects A and D with respect to the distance d = 0.5 $\sim$ 0.75 [m] by 0.05 [m], where a solid line with white circles

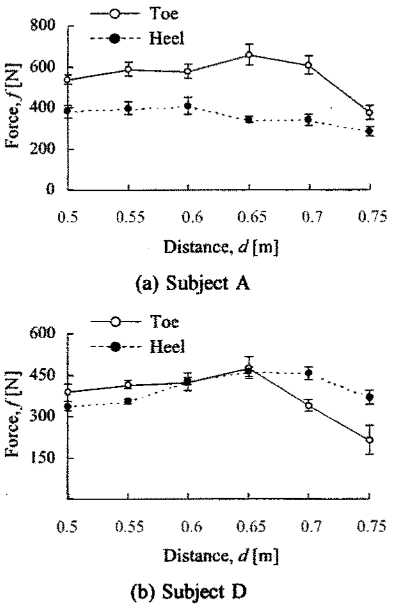


Fig. 8. The relation between end-point force by toe/heel and distance between trochanter major and lateral malleolus (Subjects A, D)

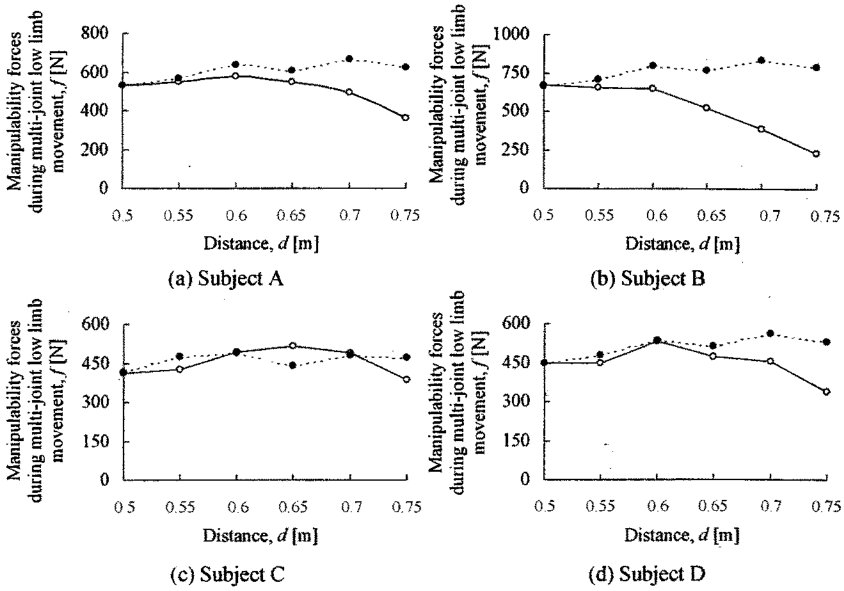


Fig. 9. The force manipulability in the leg movements according to the distance d by the proposed method and the previous method

TABLE II

LINK LENGTH OF SUBJECTS

Subject	l_1 [m]	l_2 [m]	l_3 [m]
A	0.38	0.40	0.15
B	0.41	0.39	0.14
C	0.40	0.40	0.14
D	0.41	0.41	0.15

represents the changes of the end-point force exerted by the toe while a dotted line with black circles does by the heel. The subjects generate the largest force around 0.65 [m], while it is difficult with the lower limbs fully extended at 0.75 [m]. The similar characteristics were observed on other subjects.

2) Evaluation of human movements: The analysis was conducted with the measured joint angles $\boldsymbol{\theta} = (\theta_1, \theta_2, \theta_3)$ and the measured link parameters (See Table II) in the experiments with subjects. The joint-torque activation level was set at $(\alpha_1, \alpha_2, \alpha_3) = (0.4, 0.4, 0.4)$.

Figure 9 shows the force manipulability toward the x-axis by the proposed force ellipsoid and the FME [2] with the measured leg posture, where the direction of hip extension, knee extension, and dorsiflexion were set as the positive direction. A solid line with white circles represents the predicted force by the proposed method, while a dotted line with black circles by the conventional method. In the conventional method, a large end-point force can be generated even where the lower limbs is in fully extended as shown in Fig. 9(a) against the experimental results by the subjects (See Fig. 8). On the other hand, the proposed method using human joint-torque characteristics can predict the similar tendencies with the observed results; the end-point force becomes smaller as the distance d increases.

Figure 10 illustrates the force manipulability ellipses for Subject A derived by the conventional method and the proposed method with respect to the distance d under $\theta_{chair} = 0$ [rad] (See Fig. 5). In the conventional method, the force manipulability toward the x-axis can easily increase 1.5 times as large as that under the initial posture by small changes of the leg posture, although the end-point force does not change so much in the experimental results. Consequently, it can be said that the proposed method using human joint-torque characteristics can provide more appropriate evaluation for human movements than the conventional method.

Although it is necessary to do further verification, the effectiveness of the proposed method can be verified through the presented experimental results.

3.4. Evaluation of leg postures

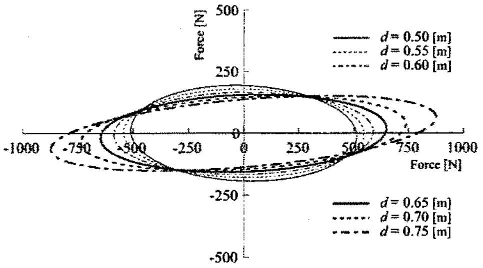
Finally, the burden level of leg joints for generating the end-point force $\boldsymbol{f}^*$ at the posture $\boldsymbol{\theta}$ was examined by using the following cost function E with $\boldsymbol{\alpha}$ in (4) as:

$$E = \boldsymbol{\alpha}^T \boldsymbol{\alpha} \geq 0. \tag{9}$$

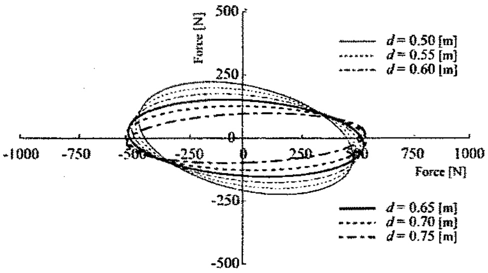
It should be noted that the joint burden level is small as the cost function E is small.

Evaluations with the end-point force $\boldsymbol{f}^*$ measured in the initial posture (d = 0.5 [m]) were carried out for the leg movements. The leg postures were calculated from the distance d and the link length shown in Table II under the foot angle θ_{foot} was at 0.873 [rad] and the chair angle θ_{chair} at 0.916 [rad]. The rotational direction of each joint was determined from the sign of each element of $\boldsymbol{J}(\boldsymbol{\theta})\boldsymbol{f}^*$.

Figure 11 shows the results for Subjects A and D. It can be seen that there exists a unique minimum value in the burden level around d = 0.70 [m]. The results indicate that the subjects feel the difficulty to generate the target end-point force $\boldsymbol{f}^*$ when the distance d is too small or too large.

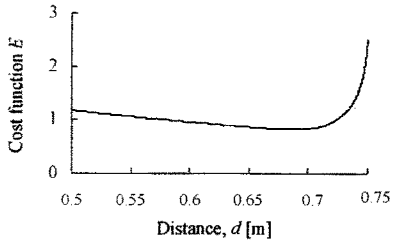


(a) The previous method

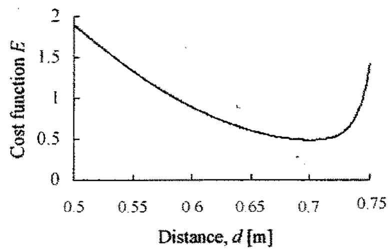


(b) The proposed method

Fig. 10. Force manipulability ellipses by the proposed method and the previous method [2] on Subject A



(a) Subject A



(b) Subject D

Fig. 11. Change of the cost function E for the distance between the end-point and trochanter major

Paradoxically there is an optimal posture in human multi-joint movements. Solving the optimization problem which minimizes the cost function E within the range of joint motion, it can be expected to obtain the optimal posture with small joint burden for the given task in consideration to the redundancy of human body.

4. Conclusions

This paper proposed a method for the quantitative analysis of human multi-joint movements. The effectiveness and usefulness of the proposed method were then verified through a series of the experimental and simulated results.

The proposed ellipsoid, however, cannot represent the accurate force manipulability toward an arbitrary direction, because it does not consider the differences of joint-torque characteristics on joint rotational directions. Future research will be directed to develop a method considering the joint rotational direction, and to investigate the influence of muscle contraction level on joint-torque characteristics in order to improve the proposed methodology.

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