Proceedings of the 2007 IEEE
International Conference on Mechatronics and Automation
August 5 - 8, 2007, Harbin, China

Determination of Human Motion for Rehabilitation Based on Time-Scale Transformation

Satoshi Tanaka[1, 2], Takahiro Wada[2] and Kosuke Kawakita[2]
[1] *Department of Physical Therapy, Prefectural University of Hiroshima*
Gakuen-machi, Mihara, Hiroshima, Japan
[2] *Department of Intelligent Mechanical Systems, Kagawa University*
Takamatsu, Kagawa, Japan
s-tanaka@pu-hiroshima.ac.jp, wada@eng.kagawa-u.ac.jp

Abstract – **This paper presents a method to determine human motion for upper limb rehabilitation. In the rehabilitation training, appropriate load should be given to each part of the human body such as joints and muscles based on the patient's conditions and symptom. Such load is changed according to the kinematic and dynamic properties of multi-link body motion. On the other hand, it is difficult for human to execute complex motion even if optimal motion trajectory for rehabilitation is instructed precisely. In this paper, we explore a method to determine human rehabilitation motion to realize appropriate load by understanding the characteristics of human body dynamics with application of the time-scale transformation characteristics of dynamics.**

Index Terms – Human motion, rehabilitation, dynamics, time-scale transformation.

I. Introduction

In rehabilitation training such as muscle strength exercise, appropriate load should be given to each part of human body such as muscles, joints according to trainee's conditions and symptom. Such load is changed according to the motion patterns as well as the body posture. Physical therapists determine appropriate body motion pattern from the given symptom and conditions of trainees and/or medical doctors diagnosis results based on their experiences. Thus, a systematic method to determine human motion based on the trainee's condition is required for the field of rehabilitation to obtain more effective training in more complex situations. For trainees with serious symptom, passive motion such as continuous passive motion (CPM) equipment is utilized [1]. On the other hand, for trainees with relatively light symptom, active motion is effective. For active motion, determination and instruction of motion is important to get appropriate load according to patient's conditions.

To assess load exerted on the human body, solving dynamics problem plays important roles. In motion analysis of body motion, motion capture system with cameras and markers is utilized to measure joint motions, then joint torque is estimated according to the given body motion by solving inverse dynamics problem. In this method, trainee's body movement in interests is required to calculate joint load. Thus, for obtaining optimal trajectory, many body movements have to be performed and measured to be compared. On the other hand, body motion can be calculated by specifying joint torque in time series pattern by solving forward dynamics problem. It is difficult to determine time history of feasible joint motion according to trainee's condition theoretically and also it is difficult for the trainee to perform accurately the generated motion even if complex body motion is obtained with much effort. From these view points, it is important to calculate appropriate but simple human body motion for rehabilitation by specifying load exerted on trainee's muscles and joints.

On the other hand, knowing how the load such as joint torque and muscle force is affected by change of joint motion is helpful to understand how appropriate human motion for rehabilitation should be derived. For example, in rehabilitation exercise, similar motion is employed and sometimes only the motion speed is changed because it is easy for trainee. Suppose that the spatial trajectory of the joint does not changed but linearly changed in time scale. On the other hand, based on the structure of dynamic equation, change of the joint torque according to the speed change can be analytically calculated just multiplying a term of the dynamic equation by scaling factor of the time change if only the base motion can be calculated in advance[2]. In this paper, we explore a method to determine human rehabilitation motion to realize appropriate load by applying the time-scale transformation characteristics of dynamic equation.

II. Concept of Generation of Rehabilitation Motion

In muscle strength exercise, it is important to generate body motion that can realize appropriate activation/load in target muscle. It is difficult to generate time series pattern of joint motions that is feasible. In addition, even if we have the optimal motion precisely, it may be impossible for trainee to execute the given motion due to complexity of the motion sequence etc.

In this paper, we deal with rehabilitation motion for an upper limb. Fig.1 illustrates a flow of rehabilitation that we suppose in this paper. Based on the symptom, target quantity of load of muscle is given. Then the base motion is determined based on determination of posture and spatial trajectory. This determination of base motion might be realized with systematic manner with the characteristics of the dynamic equation or with the experiences of therapists and the measurement in the real environment. Also, suppose that inertia property and kinematic property and other information are given to build a dynamic equation of motion. From the dynamics and joint angles of base motion, joint torque by base motion is calculated. Based on time-scale transformation, change of joint torque is easily calculated for any given speed of the motion. Based on the dynamics, speed of the motion can be determined that realize target load by changing speed of the motion. Then, only speed of the motion is instructed for the trainee. It is easier to realize the given motion rather than specifying the detailed joint angle profile in time.

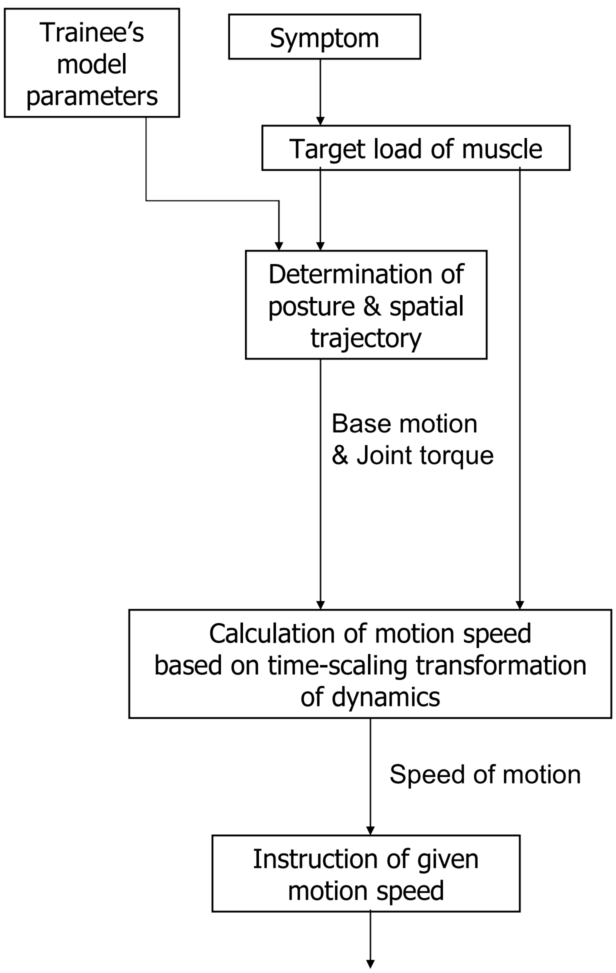


Fig. 1 Flow of assumed rehabilitation method.

III. Calculation of Change of Joint Torque according to Change of Motion Speed

A. Effect of Speed Change on Joint Torque

Assume that human body is modeled as link segment model in Lagrangian form in (1) [2][3].

$$R(q(t))\frac{d^2q(t)}{dt^2}+\left(\frac{dq(t)}{dt}\right)^T S(q(t))\left(\frac{dq(t)}{dt}\right) + D\frac{dq(t)}{dt}+g(q(t))=\tau(t) \quad (1)$$

where $q(t)\in R^n$ is time history of joint angles of n DOF, matrix $R(q(t))\in R^{n\times n}$ denotes the inertia matrix and $S(q(t))$ is a tensor belonging to $R^{n\times n\times n}$ corresponding to centrifugal and Coriolis force, respectively. Matrix $D\in R^{n\times n}$ denotes damping in joint space. Vector $g(q(t))\in R^n$ illustrates gravity term and $\tau\in R^n$ represents the joint torque.

Assume that motion pattern in joint angles can be expressed as shown in (2) if the speed of the motion is changed.

Now suppose that motion pattern

$$q_0(t)\in R^n \quad t\in[0,T_0] \quad (2)$$

is given as the base motion and the corresponding joint torque is given by $\tau_0(t)\in R^n$, where $q_0(t)$ and $\tau_0(t)$ satisfy the (1).

Given another joint trajectory

$$q_1(t/k)=q_0(t) \quad t\in[0,T_0] \quad (3)$$

that is the same spatial trajectory but just time-scale transformed by k times in time. Here k > 1 represent higher speed than base motion. It should be noted that arbitrary time-scale transformation can be realized theoretically [2][3] but it is not effective for rehabilitation motion because it is difficult for human trainee to realize complex and accurate motion. Now it should be noted that the following equations are satisfied for the time derivative of the joint trajectories:

$$\frac{dq_0(t)}{dt}=\frac{1}{k}\frac{dq_1(t/k)}{d(t/k)} \quad (4)$$

$$\frac{d^2q_0(t)}{dt^2}=\frac{1}{k^2}\frac{d^2q_1(t/k)}{d(t/k)^2} \quad (5)$$

Replacing $q(t)$ and $\tau(t)$ to $q_1(t/k)$ and $\tau_1(t/k)$ in (1) yields (6)

$$R(q_1(t/k))\frac{d^2q_1(t/k)}{d(t/k)^2} + \left(\frac{dq_1(t/k)}{d(t/k)}\right)^T S(q_1(t/k))\left(\frac{dq_1(t/k)}{d(t/k)}\right) + D\frac{dq_1(t/k)}{d(t/k)} + g(q_1(t/k)) = \tau_1(t/k) \quad (6)$$

Substituting (4) and (5) into (6) yields the followings:

$$k^2\left\{R(q_0(t))\frac{d^2q_0(t)}{dt^2} + \left(\frac{dq_0(t)}{dt}\right)^T S(q_0(t))\left(\frac{dq_0(t)}{dt}\right)\right\} + kD\frac{dq_0(t)}{dt} + g(q_0(t)) = \tau_1(t/k) \quad (7)$$

Equation (7) can be rewritten as follows:

$$\tau_1(t/k) = [p_1(t) \quad p_2(t) \quad p_3(t)]\begin{bmatrix} k^2 \\ k \\ 1 \end{bmatrix} \quad (8)$$

where

$$\begin{aligned} p_1(t) &\equiv R(q_0(t))\frac{d^2q_0(t)}{dt^2} + \left(\frac{dq_0(t)}{dt}\right)^T S(q_0(t))\left(\frac{dq_0(t)}{dt}\right) \\ p_2(t) &\equiv D\frac{dq_0(t)}{dt} \\ p_3(t) &\equiv g(q_0(t)). \end{aligned} \quad (9)$$

Note that the torque of base motion can be calculated by substituting k=1 as follows:

$$\tau_0(t) = p_1(t) + p_2(t) + p_3(t) \quad (10)$$

Namely, if the torque pattern according to a motion trajectory is given as shown in (10), change of torque pattern by time-scale transformation can be obtained in (8). Thus, the scalar k is determined such that the appropriate torque pattern is realized. In the remaining part of this paper, damping of the joint is omitted for simplicity.

B. Determination of Speed corresponding to Limit of Joint Torque

Hollerbach [2] also have derived how to obtain the motion speed realizing torque within torque limitation.

Now, let us consider how to determine the parameter k. Suppose that the torque limitation is given by the followings according to the patient's symptom and condition;

$$L_{\min} \le p_1(t)k^2 + p_3(t) \le L_{\max} \quad \forall t, \quad (11)$$

where $L_{\min}$ and $L_{\max}$ are lower and upper limitation of the joint torque. Now we define the followings;

$$\begin{aligned} R_{\max}(t) &\equiv -p_3(t) + L_{\max} \\ R_{\min}(t) &\equiv -p_3(t) + L_{\min} \end{aligned} \quad (12)$$

Then, (11) can be rewritten with (12) as follows:

$$R_{\min}(t) \le p_1(t)k^2 \le R_{\max}(t) \quad (13)$$

As described in [2], there are there cases as (i) $R_{\min}(t) \ge 0, R_{\max}(t) \ge 0$, (ii) $R_{\min}(t) \le 0, R_{\max}(t) \ge 0$, (iii) $R_{\min}(t) \le 0, R_{\max}(t) \le 0$. In case (i) and (iii), feasibility of k is determined by the sign of $P_1(t)$ and sometimes there is no feasible solution for any k. It means that the posture of the trainee is not suitable.

On the other hand, in the case of (ii), the following solution is obtained.

$$k^2 \in \begin{cases} \left[0, \dfrac{R_{\max}(t)}{p_1(t)}\right] & \mathit{for}\ p_1(t) > 0 \\ \left[0, \dfrac{R_{\min}(t)}{p_1(t)}\right] & \mathit{for}\ p_1(t) < 0 \\ [0, \infty] & \mathit{for}\ p_1(t) = 0 \end{cases} \quad (14)$$

By calculating (14) for all through the motion time the appropriate set of scaling factor k as the solution is obtained.

From the above, it is found that $R_{\min}(t) \le 0, R_{\max}(t) \ge 0$ plays an important role to control joint torque. This means that appropriate posture of the human body should be selected so that $R_{\min}(t) \le 0, R_{\max}(t) \ge 0$ is satisfied for appropriate rehabilitation. Then, load can be controlled by changing the time scaling factor k.

IV. Examples of Joint Torque Calculation

In this section, calculation results of torque change by changing motion speed are shown using the method introduced in the previous section to know how it works. For the sake of simplicity, flexion and extension of elbow is considered remaining part of this paper.

Model parameters used in the calculation are given in Table I based on [4].

TABLE I
MODEL PARAMETERS

Length of forearm	0.2m
Position of center of gravity of arm	0.078 [m]
Mass of fore arm	0.96 [kg]

The simulated motion is as follows: arm curl from the anatomical basic posture to 90[deg] in flexion. Conditions in motion time are 0.5, 1, 1.5, 2, 2.5, and 3[s]. Assume that the trajectory of elbow angle in base condition is expressed by 5-

th order of time and six coefficients are determined with boundary conditions of angles and zero values of velocities and accelerations at $t=0$ and $t=T$ as shown in Fig.2.

Fig.3 denotes the calculation results based on (8). According to the change of motion time, say, parameter k, change of joint torque is shown. Torque in 2.5s and 3.0s conditions is almost same and the torques can be understood as gravity term $p_3(t)$. Once we have gravity term $p_3(t)$ and dynamic effect term $p_1(t)$ such as 1s condition, we can imagine the change of joint torque. For example, shortening the motion time results in large peak of the torque.

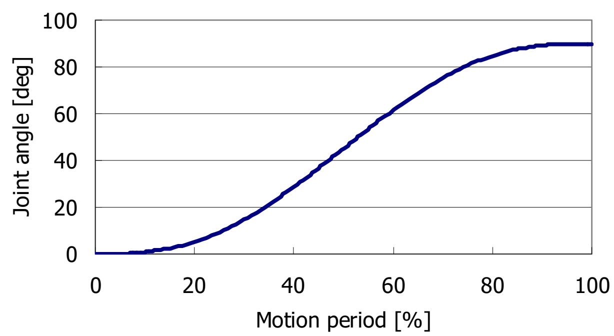


Fig. 2 Time history of joint angle.

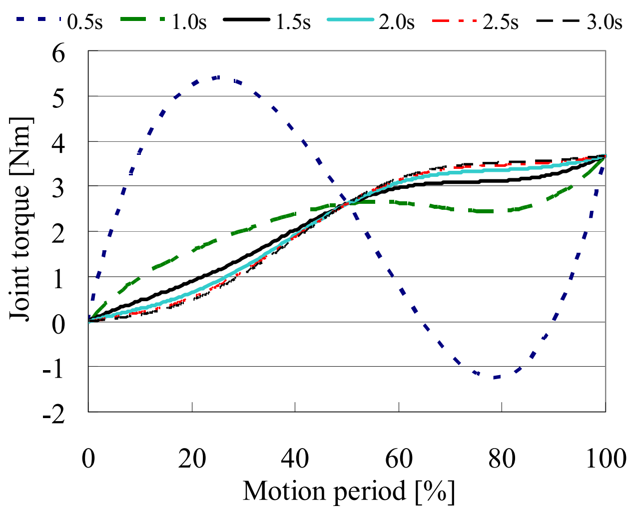


Fig. 3 Change of Joint Torque.

As mentioned in the previous section, determination of posture and spatial trajectory satisfying $R_{\min}(t) \le 0, R_{\max}(t) \ge 0$ is indispensable. In other words, as shown in Fig.3, forming the shape $p_3(t)$ in this plane is important by considering the direction of gravity.

V. Experiments

A. Experimental Comparison of joint trajectory

To use the calculation method, assumption of the same spatial trajectory even if the different speed is required. Thus, the joint trajectories in different speed are compared.

Two males participated in the experiment. The participants were in the stance posture with the upper limbs in the anatomical basic posture but grasped 1.5kgf weight by his right hand. The participants were instructed that from the basic posture, elbow was curled into 90[deg] in flexion in predetermined time 0.5, 1, 1.5, 2.0, 2.5, and 3.0[s] presenting by a metronome. Five trials are performed in each condition. Of course, the time series pattern of joint speed is not/ cannot be specified.

Joint angle was measured by motion capture system with cameras and markers, VICON system. EMG signals of biceps bracii and tricpes brachii are also measured to compare muscle activation in different conditions as shown in Fig.4. The EMG signal is picked up by surface electrodes that are attached according to Perotto's method [5].

Fig.5 illustrates the joint angle in 0.5, 1 and 2[s] conditions of one of the participants as examples. The horizontal axis denotes percentage of joint angle in time, that is, it is equivalent to t/T times 100, where t and T denote time and total time of motion from 0[deg] to 90[deg], respectively. Namely, 0[%] and 100[%] mean 0[deg] and 90[deg] of elbow flexion angle, respectively. Solid lines and dotted lines are mean and mean+SD/mean-SD for the five trials per condition, respectively. As shown in the figure, only small deviations are observed within condition.

Fig.6 illustrates the mean joint angle of the all conditions. The deviation between the conditions is not large and it is almost the same level that is seen in the same condition except for the condition of 0.5[s], very fast motion and 3.0[s], very slow motion. As the results, the time scale assumption can be fair by specifying only the total motion time.

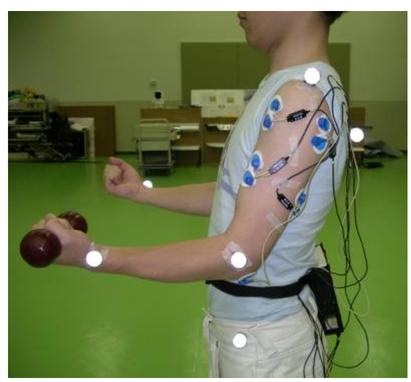
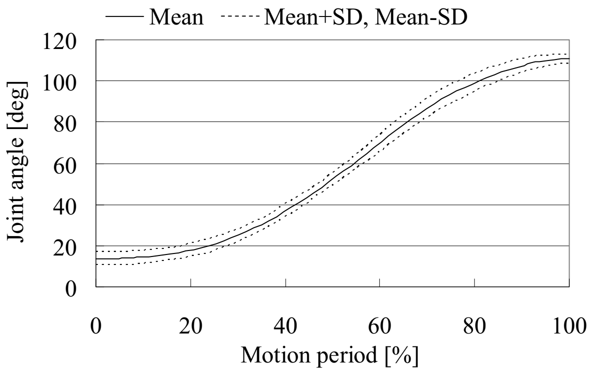
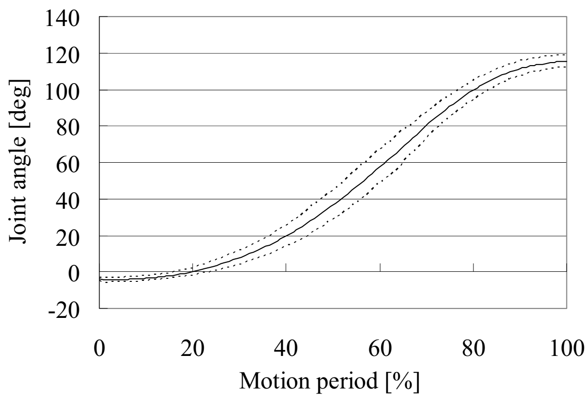


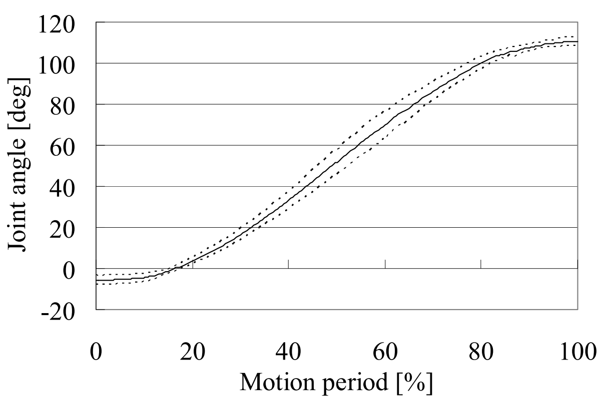
Fig. 4 Experimental Scene.



(a) 0.5 s



(b) 1.0 s



(c) 2.0 s

Fig. 5 Joint Trajectory

B. *Comparison of Joint Torque*

Fig.7 shows the validity of the proposed method, joint torque is calculated from the joint angle, that is motion pattern given in Fig.6. As shown Fig.7, faster motion period results in larger joint torque especially for the end of motion. On the other hand, torque patterns of the motion slower than 2s condition are almost same and that torque patterns can be understood as that of the gravity effect. Let us compare torque of 1.5s with that of 1.0s. From (8), the ratio of torque according to dynamic effect of 1.0s to that of 1.5s should be $(1/1.5)^2=0.44$. In the experimental results, the ratio is about 4 around the peak value of the end of the motion.

C. *Comparison of Muscle Activation*

Fig.8 shows that EMG signals of biceps brachii and triceps brachii in one trial of three conditions of 0.5s, 1.0s, and 2.0s after rectification and enveloped as example.

As shown in the figure, muscle activation is increased in faster motion. In 0.5s condition, EMG signal is changed with similar shape of torque changes in Fig.7. On the other hand, the EMG signal of 1.0 and 2.0s are not so changed even though the EMG of 1.0s is slightly larger than that of 2.0s. In the end of the motion, EMG of 0.5s is smaller than those of 1.0s and 2.0s. It shows that for slower motion leads to larger muscle activation for increasing joint rigidity. In order to take such phenomena, muscle activation model should be introduced as a future study to estimate muscle activation.

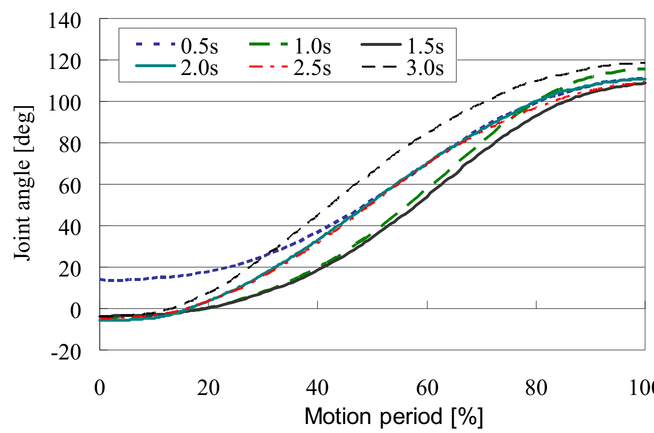


Fig. 6 Joint Trajectories for all conditions.

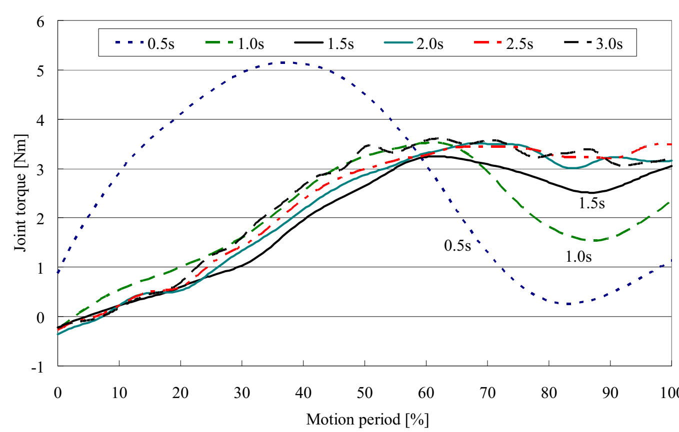
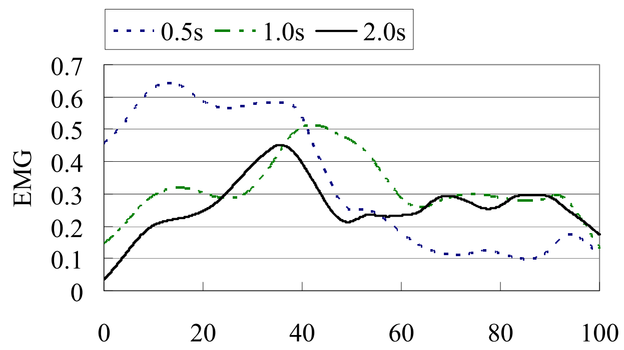
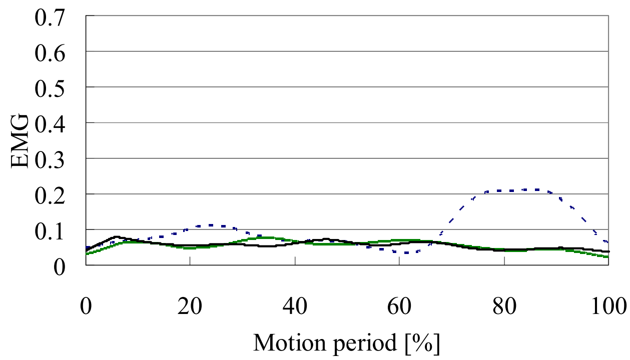


Fig. 7 Joint Torque



(a) Biceps brachii



(b) Triceps brachii

Fig. 8 Muscle Activation

VI. Conclusions

In this paper, a method to determine human motion speed for upper limb rehabilitation training was introduced to realize appropriate load for muscles based on time-scale transformation of dynamic equation. First, a method to estimate joint torque change by changing motion speed was introduced. A method to realize appropriate joint torque was introduced. In addition, we pointed out that forming torque pattern by gravity effect plays an important role to realize the given load and this lead to determination of posture of the trainee and spatial trajectory of the training motion.

From the experimental results, it is shown that body movement is changed almost linearly in time if the speed of the movement is changed in a certain range. Thus, the linear time-scale transformation can be applied to the human body to estimate joint torque. Experimental results also show that the joint torque is estimated appropriately. Furthermore, the muscle activation is measured to compare with the results of the increase of estimated joint torque. Muscle activation was increased in faster motion. In 0.5s condition, EMG is changed with similar shape of the change of torque. On the other hand, EMG signals in slower motions are not changed largely but with continuous contraction.

As the future work, we will show the validity of the introduced method in the complex motion such as multi-joint motions. In addition, we will expand the method to estimation of muscle activation. Furthermore, a method to determine the desired load for each joint and muscle according to trainee's condition will be developed by introducing muscle activation model.

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