Paper:

Upper-Limb Power-Assist Control for Agriculture Load Lifting

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This paper discusses upper-limb power-assist control using a pneumatic rotary actuator to support shoulder movement and an air cylinder to support elbow movement by agricultural workers lifting a 30 kg rice bag without inducing low back pain. Surface electromyogram (EMG) signals are used as trigger signals, joint support torque is calculated based on the antigravity term of necessary joint torque, estimated based on the dynamics of a human approximated link model. Experimental results demonstrate the effectiveness of proposed power-assist control.

Keywords: power assist system, wearable robot system, exoskeleton, pneumatic actuator, emg, elbow motion, shoulder motion, farming, lift-up motion

1. Introduction

We have focused our attention on agriculture-oriented power-assist suits for reducing the physical burden on aging farm laborers. Power-assist suits developed thus far include the robot suit HAL that uses an electric motor with a reducer [1], an exoskeleton using an electric motor with a reducer [2], a wearable standalone power-assist suit with a pneumatic rubber bellow actuator [3], a muscle suit using a pneumatic rubber muscle [4], a sheet-like curved pneumatic rubber muscle [5], an extremity exoskeleton using a hydraulic cylinder for military use [6], and a power-assist suit using an ultrasonic motor [7].

We use upper-limb power-assist aiding agricultural workers' elbows and shoulders in lifting that would otherwise strain them excessively. Using a pneumatic actuator consisting of an air cylinder and a pneumatic rotary actuator, we assist laborers in lifting 30 kg grain packs. Joints are flexible thanks to compressed-air use. Using air reduces the risk to harm human when the power-assist receives impact forces.

Muscle torque is conventionally estimated using electromyogram (EMG) signals based on neural-network learning theory, and varies widely among individuals and fatigue levels, requiring precise calibration. When an EMG signal is used as a trigger, subsequent assistance executed based on known action patterns requires a database and is interrupted if the action sequence is switched. With an EMG trigger and approximating human movement by a multijoint rigid link model, we use measured joint angle and load to statically calculate torque required to hold an object, achieving assist that does not cause discomfort to the wearer. This assist aides agricultural workers in lifting heavy objects in uses with low work speed and acceleration that maximize the effects of gravity.

Table 1. Power-assist specifications.

Joint	Elbow	Shoulder
Actuator Type	Air cylinder —	Air rotary actuator Double vane
Inner Diameter[mm]	20	50
Stroke[mm]	125	—
Mass[kg]	0.22	0.82
Torque range [Nm] (Pressure range [Mpa])	4.40〜13.2 (0.2〜0.6)	4.02〜14.3 (0.2〜0.6)
Movable range [deg]	0〜120	0〜100

2. Power-Assist

Table 1 shows power-assist specifications. One shoulder is assisted in bending and stretching in load lifting, while the other acts as a passive rotating joint with 2 degrees of freedom (2DOF). Elbow torque varies with the angle due to the link structure. **Table 1** is maximum torques at the pressures. A personal computer (PC) controller and portable compressor are planned for adding to the back, with total load to be 10 kg. **Fig. 1** gives an overview of the wearable power-assist system and **Fig. 2** shows the system configuration.

3. Power-Assist Control

Figure 3 shows power-assist control. Load is estimated using a strain gauge on the power-assist forearm support, while individual joint angles are simultaneously measured using a potentiometer to calculate posture-

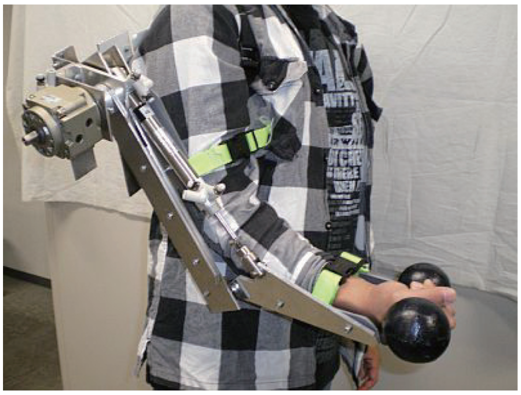


Fig. 1. Power-assist overview.

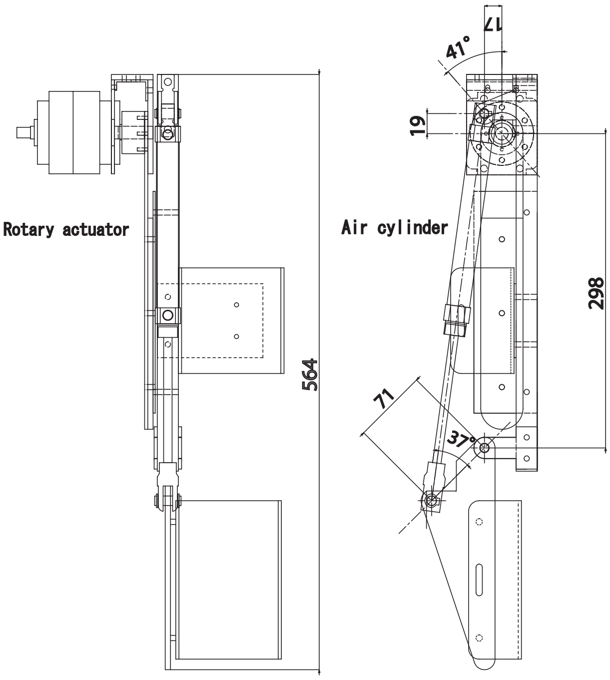


Fig. 2. Power-assist configuration.

responsive holding torque. While holding torque is being calculated, EMG signals determine whether elbows and shoulders are bent or stretched in response to measured muscle movement, and action torque is added as needed.

This control enables torque output to assist the wearer, regardless of actions, such as dropping an object. The controller laptop PC uses a 1.2 GHz CPU and 512 MB of memory. A 16-bit A/D converter uses an input range of ±10 V to provide the controller with EMG information, a potentiometer for detecting the joint angle, and a strain-gauge voltage-signal sensor. Control output uses a 16-bit D/A converter with an output range of ±10 V and a control period of 10 ms.

3.1. EMG Measurement

Electromyogram (EMG) is an electric potential generated by muscle contraction. We use minimally invasive skin-surface dry-triode EMG electrodes. A high-pass filter attenuates frequency band signals ≤10 Hz and a hum noise filter that attenuates frequency band signals at 50 Hz and 60 Hz to amplify signals 1000-fold before measuring them.

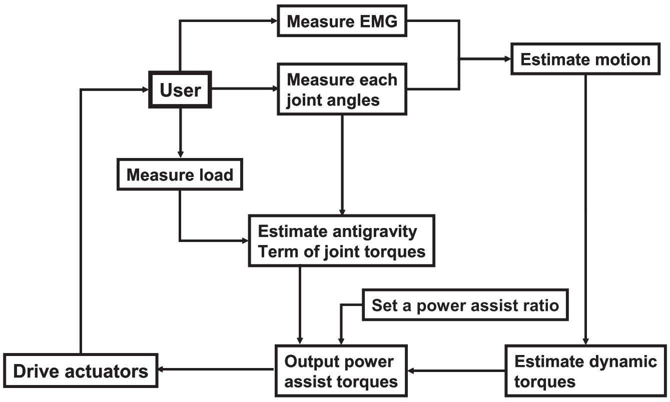


Fig. 3. Control.

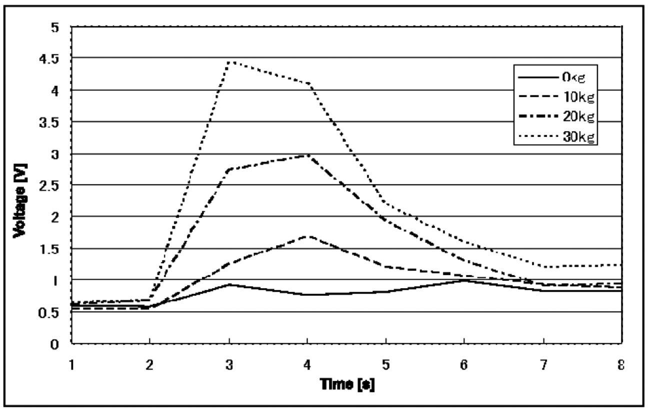


Fig. 4. RMS strain signals.

3.2. Load Estimation

Load was estimated using a strain gauge on the power-assist forearm support for the elbow and measuring strain gauge output voltage at different loads – no load, and load lifting 10, 20, and 30 kg bags 0.6 m from the ground every 10 ms. The subject lifts and holds up a bag for 2 sec, then releases it after 1 sec. The strain gauge amplifier was set to zero when the power-assist was worn without a load.

Voltage varies too much to extract the trend of the load, so we used root mean square (RMS) processing on the intensity of variable values to extract feature quantities as follows.

where N is the number of segments and v_i the i th sampling voltage:

$$RMS = \sqrt{\frac{1}{N}\sum_{i=0}^{N} v_i^2} \quad \ldots\ldots\ldots\ldots \quad (1)$$

We set the number N of segments to 100 and RMS signals during 1000 ms. **Fig. 4** shows RMS processing result, where the solid line is the result with no load (0 kg), the dashed line the result for a 10 kg load, the dot-dash line that for a 20 kg load, and the dotted line that for a 30 kg

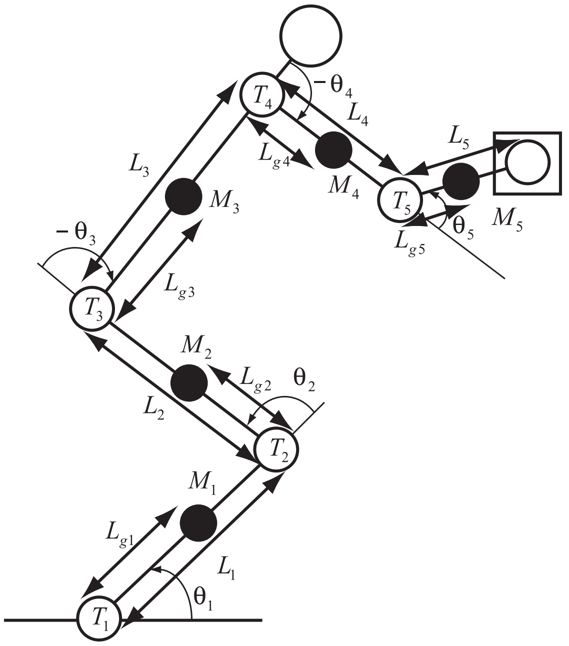
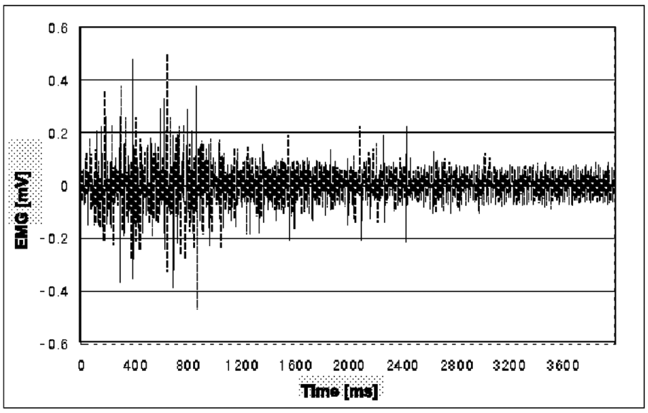


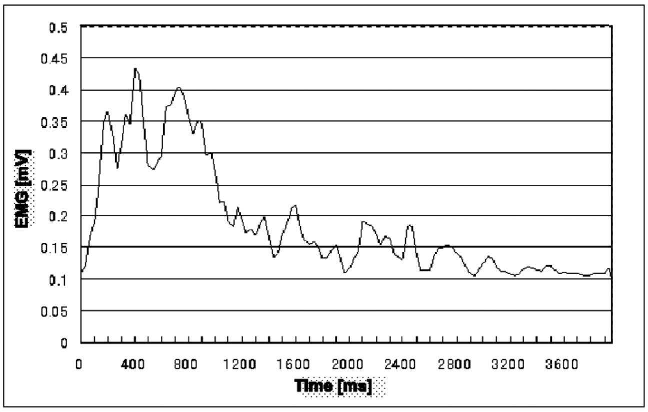
Fig. 5. Simple 5-link model.

Table 2. Link parameters.

Joint No.	1	2	3	4	5
Joint Torque[Nm]	T_1	T_2	T_3	T_4	T_5
Mass [kg]	M_1	M_2	M_3	M_4	M_5
Link Length [m]	L_1	L_2	L_3	L_4	L_5
Length to COG [m]	L_{g1}	L_{g2}	L_{g3}	L_{g4}	L_{g5}
Joint Angle [deg]	θ_1	θ_2	θ_3	θ_4	θ_5



(a) Raw EMG signal



(b) RMS EMG signal

Fig. 6. EMG signals for biceps brachii.

load. **Fig. 4** shows the load-dependent voltage change patterns used to estimate load.

3.3. Holding Torques Calculation

The human musculoskeleton's redundancy makes it difficult to express joint torque generated by muscle expansion and contraction in equations. We modeled the musculoskeleton using a multijoint rigid link mechanism to develop the expression for joint torque, as shown in **Fig. 5**. Eq. (2) expresses joint torque derived from the model.

Eq. (2) parameters are listed in **Table 2**. Posture-responsive holding torque on each joint is calculated by substituting joint angles into Eq. (2) given the length of each part and mass of the wearer measured in advance and, at M_5, the mass of the load to be lifted estimated for other than the mass of the wearer's forearm. This holding torque is used for power-assist.

3.4. Trigger Signals Movement

The wearer's movement is important to power-assist control from the aspect of comfort. We use EMG signals to estimate wearer's movement, measuring elbow bending and stretching through the biceps and triceps brachii muscle and shoulder bending and stretching through the anterior and posterior deltoids.

EMG signals are vulnerable to noise, making it difficult to use them directly, so we use feature extraction by RMS on obtained EMG signals. Eq. for the processing is the same as Eq. (1).

Figure 6 (a) shows raw EMG signals and **(b)** RMS EMG signals for 200 ms with the number N of segments set to 20 for the biceps brachii muscle when the elbow is bent for 1 sec without holding anything, pauses for 1 s, and stretches for another 1 s. EMG signals for the biceps brachii muscle become greatest while the elbow is bending. Since RMS EMG signals became particularly great, we estimated action with these signals as muscle activity trigger signals when the amount of change during 200 ms was $\geq$0.2 mV. We obtained similar results in shoulder bending and stretching.

$$\left.\begin{aligned}
T_5 &= M_5 L_{g5} g \cos(\theta_1+\theta_2+\theta_3+\theta_4+\theta_5) \\
T_4 &= (M_4 L_{g4} + M_5 L_4) g \cos(\theta_1+\theta_2+\theta_3+\theta_4) + T_5 \\
T_3 &= \{M_3 L_{g3} + (M_4+M_5)L3\} g \cos(\theta_1+\theta_2+\theta_3) + T_4 \\
T_2 &= \{M_2 L_{g2} + (M_3+M_4+M_5)L_2\} g \cos(\theta_1+\theta_2) + T_3 \\
T_1 &= \{M_1 L_{g1} + (M_2+M_3+M_4+M_5)L_1\} g \cos\theta_1 + T_2
\end{aligned}\right\} \quad \cdots\cdots (2)$$

Table 3. Experimental parameters.

Joint No.	1	2	3	4	5
Mass M[kg]	5.33	8.70	18.8	2.55	2.33
Link Length L[m]	0.44	0.49	0.42	0.30	0.25
Length to COG Lg[m]	0.22	0.25	0.21	0.15	0.13

4. Experiments

4.1. Objectives

In agricultural work, load lifting puts a heavy strain on workers, especially in work such as loading and unloading 30 kg rice bags using wheelbarrows. We verify the feasibility of our power-assist by lifting a rice bag from a 0.6 m shelf – the same height as a wheelbarrow – using the power-assist. We verify that appropriate assist torque is been output and that RMS EMG signals for the biceps and triceps brachii muscle and the posterior and anterior deltoids function as triggers. EMG measurement signals with and without power-assist are compared to verify that assistance is actually achieved.

4.2. Methods

The test subject, a 75 kg male adult, wears the assist mechanism and lifts a 30 kg rice bag 0.6 m high, the same height as shelves for holding rice bags. We measure joint angle and assist torque from the time the rice bag is lifted to when it is put down, voltage from the load-measurement strain gauge, and RMS EMG signals, and compare EMG signals with and without power-assist used. For each of the link parameters in **Table 2**, we substitute the values in **Table 3** into Eq. (2), measuring only elbow angle θ_5 and shoulder angle θ_4. Assuming the subject is upright, we calculate holding torque using ankle angle θ_1 of 70°, knee angle θ_2 of 35°, and torso angle θ_3 of $-40°$ when the power-assist wearer lifts the bag.

4.3. Load Lifting Confirmation

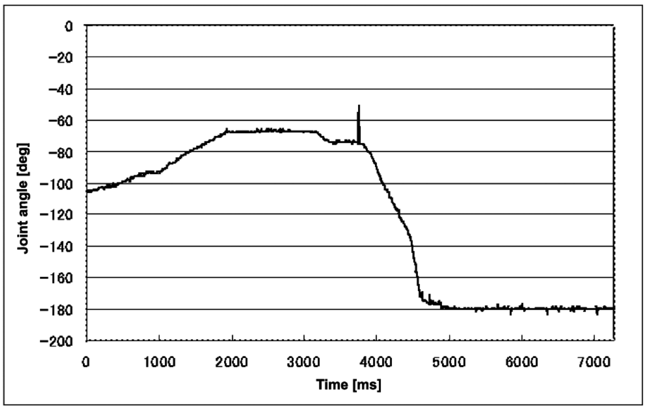
Figure 7 shows the power-assist wearer lifting the rice bag grasping the bag, lifting onto the 0.6 m table in 2 s, raising it into the air for 1 s, then putting it on the table in 2 s. This is confirms a very slow action because speed and acceleration are low and therefore confirms only the influence of gravity as detailed below.

4.4. Joint Angle Detection

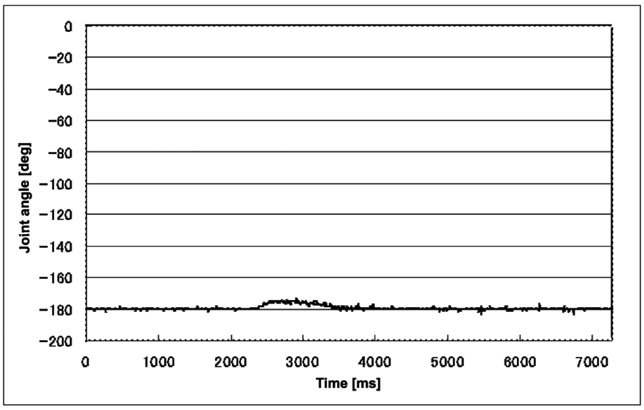
Figure 8 shows measurement results for **(a)** the change in the elbow angle and **(b)** the change in the shoulder angle.

The elbow angle is negative in the direction of stretch, increasing while the elbow is bent for the first 2 sec and remaining constant during the next 1 sec with the bag suspended, then decreasing as the elbow is stretched, although a moment occurs in which the elbow angle increases due to noise. The shoulder angle follows a similar trend. The power-assist wearer lifted the bag to a height of 0.6 m using his arms and low back.

Fig. 7. Load lifting.



(a) Elbow angle



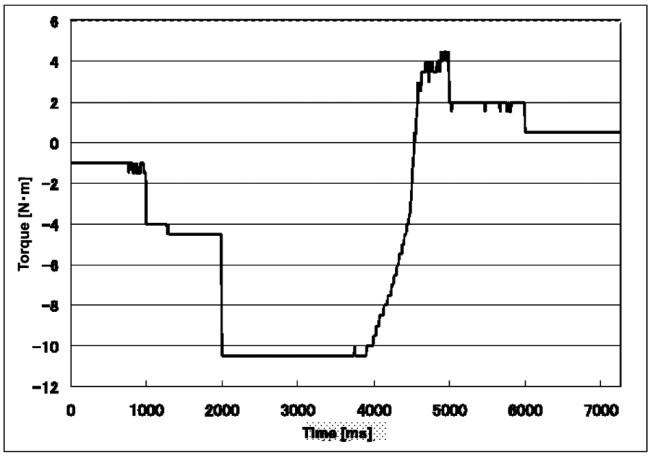
(b) Shoulder angle

Fig. 8. Joint angles measured during load lifting.

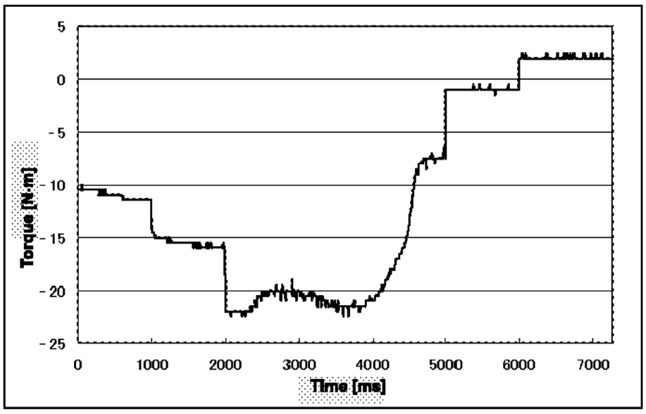
4.5. Assist Torque Verification

Figure 9 shows torque output by the power-assist during load lifting in **Fig. 7**. **Fig. 9 (a)** shows elbow assist torque and **Fig. 9 (b)** shoulder assist torque, both positive in the direction of bending and negative in the direction of stretching. Elbow assist torque output in experiments was half that of elbow holding torque T_5 obtained by Eq. (2), with shoulder assist torque output and shoulder holding torque T_4 following the same trend.

Elbow inertia torque is approximated to be about 0.005

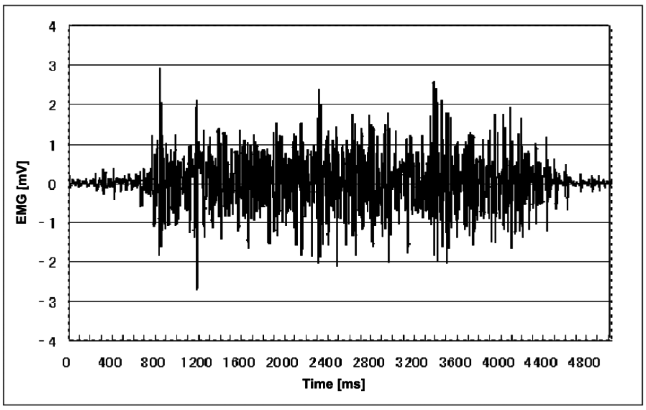


(a) Elbow torque

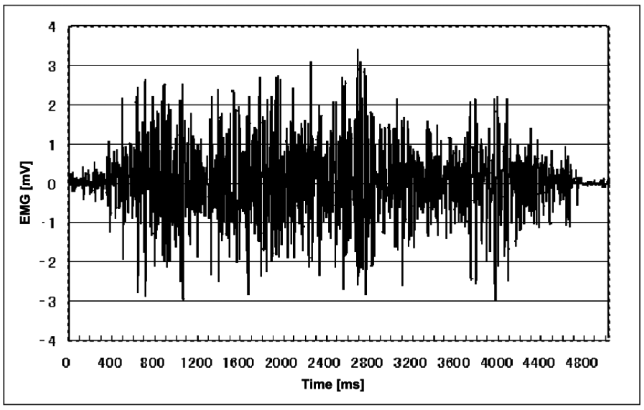


(b) Shoulder torque

Fig. 9. Power-assisting torques during load lifting.



(a) With assist



(b) Without assist

Fig. 10. Raw EMG signals for biceps brachii.

times the elbow holding torque. Since it is significantly small, we used it practically as 0.005 times the holding torque. Elbow movement was measured using EMG change as the movement trigger. The whole elbow assist torque was calculated as the addition of inertia torque and half of the statically required holding torque. We operated the actuator's target differential pressure required to generate calculated assist torque to control assist torque required for necessary the power-assist wearer's elbow.

Inertia torque of the elbow is expressed as follows:

$$\tau_5 = M_5 L_{g5}^2 a_5 \quad \dots\dots\dots\dots\dots \quad (3)$$

Elbow inertia torque τ_5 was therefore set to be 0.10 Nm, given that mass M_5 of the forearm and load is 17.3 kg, length L_{g5} to the arm's center of gravity is 0.13 m, and angular acceleration a_5 is 0.35 rad/s^2. Elbow holding torque was calculated at 21 Nm by T_5 from Eq. (2), given that the elbow angle θ_5 is 113°, and shoulder angle $\theta_4 - 180°$. Elbow inertia torque was negligible at 0.005 times the elbow holding torque.

Shoulder measurements were calculated similarly, and we found that assist torque is output in response to elbow and shoulder movement and that movement is smooth. Much the same was true for elbow and shoulder centrifugal and Coriolis force, which were skipped for practical reasons. We calculated assist torque assuming that the lower limb joint angle is constant. Assist effect sufficient for practical use was obtained from the EMG signal decrease detailed below.

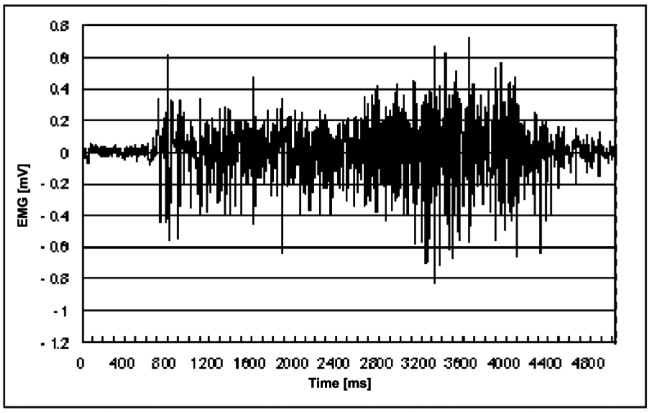
4.6. Trigger Signal Movement Verification

EMG signals for the biceps brachii muscle are used as trigger signals when the elbow is bent, (as shown in **Fig. 6**), and the same follows for EMG signals for the anterior deltoid when the shoulder is bent. EMG signals for the triceps brachii muscle and posterior deltoid, however, are difficult to use as triggers because the lack of strain generates only low-level signals when the elbow and shoulder are stretched.

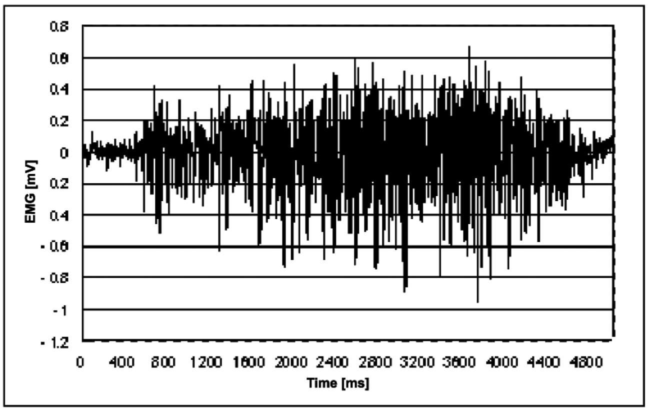
If few EMG signals are output and movement trigger signals cannot be calculated, change signals for individual joint angles are used for movement change, treating increased elbow angles as bending, decreased elbow angles as stretching, and unchanged elbow angles as holding in the air, with shoulder angles treated the same way, to disappear faltered movement at start-up and minimize wearer discomfort.

4.7. Assist Effect Verification

EMG signals output with and without assist were compared to verify power-assist effects. **Fig. 10 (a)** shows raw EMG signals for the biceps brachii muscle output with



(a) With assist



(b) Without assist

Fig. 11. Raw EMG signals for anterior deltoideus.

power-assist and **(b)** those without. **Fig. 11** shows the same for the anterior deltoid.

The comparison confirmed that EMG signals for the "antigravity" biceps brachii muscle and anterior deltoid were smaller with power-assist than without, demonstrating positive power-assist effects. EMG signals for the triceps brachii muscle and the posterior deltoid were not compared because they required no assistance and the wearer can put objects down with little strain.

5. Conclusions

We have proposed upper-limb power-assist control for use in lifting agricultural load. Assuming that the user is a multijoint rigid link mechanism, we calculated joint holding torque and proposed appropriate assistance regardless of load and posture changes.

Power-assist control requires that the wearer's movement be predicted to minimize discomfort in power-assist use, so we used EMG signals as trigger signals and measured joint angles to predict wearer movement. Experiments demonstrated that assist torque was responsive to load lifting and that assist effects actually occurred. We plan to verify assist effect and safety directly in the elderly to develop total power-assist control also for the lower limbs.

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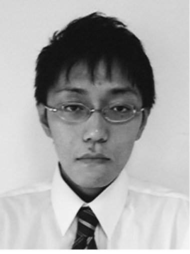

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