Power Assist Control for Leg with HAL-3 Based on Virtual Torque and Impedance Adjustment

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Abstract— **This paper describes a method of power assist control for lower body based on neuromuscular signal, s-EMG (surface ElectroMyogram/Myoelectricity) , and impedance adjustment around knee joint with the assist system, HAL (Hybrid Assistive Leg) -3 we have developed. Virtual Torque calculated by s-EMG enabled the HAL-3 to be operated as to intention of the experimental subject put on HAL, and assist the motion of lower body by predicting the moment around joints. Besides, the operator was able to swing the leg lighter by reducing the inertia and viscous friction around joint of the subject and HAL-3. In order to verify the proposed method, experiments for simple motion was performed with impedance values found by parameter identification with RLS (Recursive Least Square) method. The evaluation of assisted motion was done by Assist Efficiency (*AE*) calculated from s-EMG in nearly proportion to the operator's muscle force. The results showed the response of operational signal into actuator with impedance adjustment was improved dramatically, and the amplitudes of s-EMG were reduced significantly, then we could confirm the availability of impedance adjustment.**

Keywords— **HAL-3, Virtual Torque, impedance adjustment, Assist Efficiency(*AE*).**

I. Introduction

Nursing care and rehabilitation are required to be improved in accordance with aging in several country. It is important to enable physically weak person i.e., the old and the disabled to take care of themselves in that society. From the point of view of locomotion, the spheres of the disabled person who have some disorders like as neuromuscular diseases, or the aged person who have muscular atrophy are restricted in spite of using wheelchair due to stairway or unlevel ground, and it is desirable for such people to walk by themselves with respect to their requirement to move, burden of caregiver, and effectiveness of rehabilitation. Nevertheless, only few attempts have been made at another device to assist the leg's movement for such people. By the way, recent progress of robotics technology brings a lot of benefits in many other fields like welfare, medicine as well as in the industry. In particular, integrating humans and robotic machine into one system offer the opportunities for creating new assistive technologies that can be used in such fields [1], [2]. We have developed the s-EMG based exoskeleton system for lower body, HAL in recent years for the context as mentioned above [3], [4], [5], [6]. The power assist control of HAL has been performed essentially based on s-EMG (surface-ElectroMyogram, / Myoelectricity) of flexor and extensor muscle. HAL can cognize the operator's condition by the sensors, and enables the operator (experimental subject put on HAL) do a motion such as walking to be assisted by transferring the intention of operator to motivity of actuator through s-EMG. Our objective in this research is to realize the more effective assisted motion based on s-EMG with the impedance adjustment around knee joint. This paper describes the hardware configuration of HAL-3 in Section II, the control method with s-EMG and impedance adjustment based on estimated parameters in Section III, the evaluation method derived from s-EMG and experimental results to verify the proposed approach in Section IV.

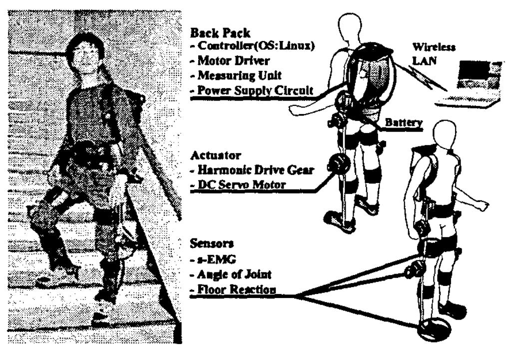

Fig. 1. HAL (Hybrid Assistive Leg) -3

II. HAL (Hybrid Assistive Leg) -3

Fig. 1. shows the system overview of the exoskeleton type power assist system, HAL (Hybrid Assistive Leg) -3 we developed, and Fig. 2. indicates the system configuration [7]. All devices for control of HAL-3 such as CPU board (PentiumII 566MHz, PCI665VRE) , motor driver (TITECH Driver PC-0121-1, 750W, ± 38V, ±8A) , measuring unit, and power source (battery, actuator: 9V × 3, controller: 10V × 1) are contained in

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backpack. A experimenter can adjust the experimental settings and monitor the operator's condition constantly with palm-top computer through wireless LAN. Besides, HAL can recognize the operator's condition from sensors installed on main body of HAL-3 for detecting the s-EMG, angle of joint, and foot-grounding and work automatically along to operator's conditions. So HAL can be operated alone without external operational environment. Harmonic drive gear and DC servo motor (Maxon, 150W, 24V) are adopted as the actuators of HAL, which have sufficient motivity to support calculated on the basis of the maximum torque in standing up motion and angular velocity in walking motion (Table I) .

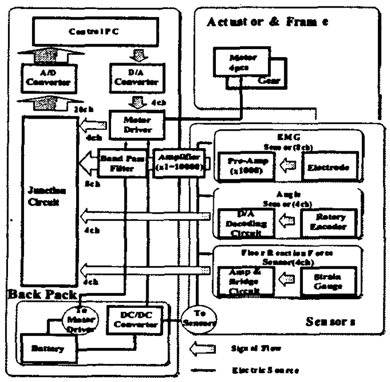

Fig. 2. System configuration

TABLE I
DATA OF KNEE AND HIP JOINT BASED ON MOTION ANALYSIS

	maximum angle $[rad]$	maximum angular velocity $[rad/sec]$	maximum torque $[kgf \cdot cm]$
hip joint	2.2	3.8	700
knee joint	2.5	5.0	800

III. Control Method

A. Virtual Torque

The predictive power assist control with operator's intention to move can be realized by Virtual Torque with adequately processed s-EMG (surface ElectroMyoGram / Myoelectricity) signal. The s-EMG is electric signal that is generated on the surface of muscle through nerve cells just after the controller of human (Central Nerve System) convey motor commands to muscle [8]. And the torque around the human's joint can be estimated by measuring the s-EMG because its amplitude is nearly in proportion to the force generated by muscle contraction. The procedure to detect the s-EMG and the relationship between myosignal and muscle force measured by force sensor are shown in Fig. 3. So the necessary the timing of joint's movement and muscle power can be calculated indirectly and previously.

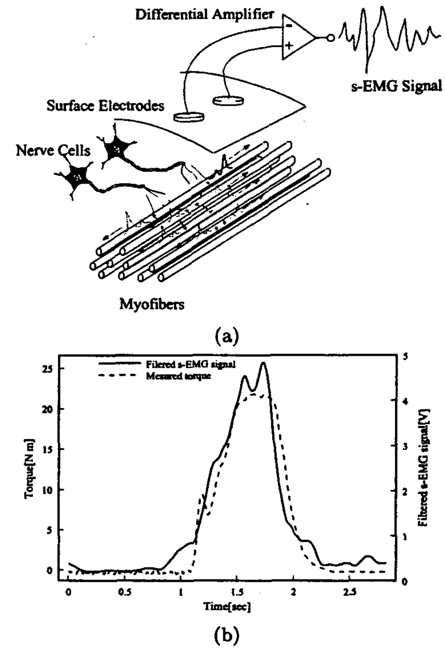

Fig. 3. (a)The procedure to generate the myosignal (b)the relationship between s-EMG and muscle force

As s-EMG as it is inappropriate to use itself for control, the raw signal is filtered through RMS (Root Mean Square) filtering process after amplified and filtered through the analog circuit [5].

$$RMS(r(t)) = \left\{ \frac{1}{T} \int_0^T r^2(t)dt \right\}^{\frac{1}{2}} \qquad (1)$$

where $r(t)$ is raw signal of s-EMG, and T is integral period. Then, the control of HAL with operator's animus to move can be performed by converting the myosignal balance of flexor muscle and extensor muscle to operational signal of actuator. Actually, estimated torque around joint, called Virtual Torque [3], is found as follows

$$\tau_{virtual}(t) = K_f E_{flx}(t) - K_e E_{ext}(t) \qquad (2)$$

where

$\tau_{virtual}(t)$ Virtual Torque;

K_f, K_e Conversion factor from s-EMG to torque;

$E_{flx}(t)$, $E_{ext}(t)$ Filtered signal of s-EMG at flexor and extensor.

The conversion factors were determined by trial and error in former research, and now, these are also decided automatically through simple motion by Neural Network or Recursive Least Square algorithm [6], [7].

B. Impedance Adjustment Around Knee Joint

The harmonic drive gear and DC servo motor with relatively high torque-to-weight ratio is adopted as the actuator of HAL, but it cannot be operated with flexibility in an ordinary way. Then, we regard that it is possible to perform the more effective assisting control if the actuator can regulate the characteristics around its joints according to a motion as human enables his/her joints to be flexible or stiff by adjusting the strain of muscle. If muscle force around knee joint is not generated, and external force works to same direction as torque of actuator, the lower thigh of operator put on HAL-3 can be represented by 1link pendulum model. The motion equation is expressed in this case as equation(3),

$$I\frac{d^2\theta}{dt^2} + D\frac{d\theta}{dt} + C\left(\theta, \frac{d\theta}{dt}\right) = \tau + J^T F_e \qquad (3)$$

where I and D are inertia and viscous coefficient around knee joint respectively, $C(\theta, \frac{d\theta}{dt})$ is non-linear term including gravity and coulomb friction, τ is torque of actuator, and F_e is external force. When the target impedance, M, B, K found based on operator's condition in contacting the environment, and target posture is set, the motion equation can be expressed by equation(4)

$$J^T F_e = M\frac{d^2\theta}{dt^2} + B\frac{d\theta}{dt} + K(\theta - \theta_0) \qquad (4)$$

where

M, B, K target value of inertia, viscous coefficient, elastic coefficient;

θ_0 angle of joint in target posture.

In order to regulate the impedance, compensation torque generated by actuator is determined as

$$\tau_{com} = (I - M)\frac{d^2\theta}{dt^2} + (D - B)\frac{d\theta}{dt} + K(\theta_0 - \theta) + C\left(\theta, \frac{d\theta}{dt}\right) \qquad (5)$$

Hence, Virtual Torque with impedance adjustment around knee joint is decided as follows

$$\begin{aligned}\tau_{sum} &= \tau_{virtual} + \tau_{com} \\ &= K_f E_{flx} - K_e E_{ext} + (I - M)\frac{d^2\theta}{dt^2} + (D - B)\frac{d\theta}{dt} \\ &\quad + K(\theta_0 - \theta) + C(\theta, \frac{d\theta}{dt}) \qquad (6)\end{aligned}$$

Fig. 4. shows block diagram of these process. The power assist control according to operator's intention with variable impedance, inertia, viscous friction, stiffness, can be performed by feedback of detected angle of joint, and calculated angular velocity and acceleration. Non-linear term is omitted in this figure.

IV. Experiment

A. Performance Indices

As amplitude of s-EMG is nearly commensurate with the torque around joint, the assisted motion by HAL-3 can be evaluated with form as

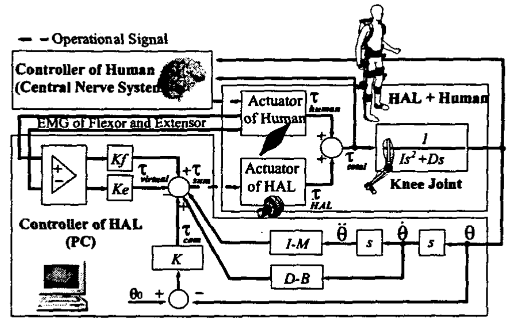

Fig. 4. Virtual Torque with impedance adjustment around knee joint

$$AE = \frac{EMG_{avH} - EMG_{avA}}{EMG_{avH}} \times 100 \qquad (7)$$

$$EMG_{avA} = \frac{1}{T}\int_0^T EMG_{Assist}(t)dt \qquad (8)$$

$$EMG_{avH} = \frac{1}{T}\int_0^T EMG_{Human}(t)dt \qquad (9)$$

where

AE Assist Efficiency[%];

EMG_{avA} average of s-EMG (with assist by HAL) ;

EMG_{avH} average of s-EMG (without assist by HAL);

$EMG_{Assist}(t)$ s-EMG (with assist by HAL) ;

$EMG_{Human}(t)$ s-EMG signal (without assist by HAL);

T measuring time.

Assist Efficiency is considered as the case of flexor and extensor separately. that is,

AE_{fl} Assist Efficiency of flexor muscle;

AE_{ex} Assist Efficiency of extensor muscle.

e.g., AE_{fl} = 60% means that HAL generates 60% of torque required to perform the same motion human have done without assist of HAL by flexor muscle.

B. Parameter Identification around Knee Joint with Recursive Least Square (RLS) Method

Parameters around knee joint must be acquired in order to apply the control method mentioned above. RLS method is adopted for parameter estimation in this research. So inertia, viscous coefficient, and mass of joint can be calculated easily all at once (In this paper, mass of lower thigh doesn't be used to control) . Discrete transfer function of operator's lower thigh is as following form [9].

$$\frac{\theta(z)}{\tau(z)} = \frac{T^2/2I\left(z^{-1} + z^{-2}\right)}{1 + (DT/I - 2)\, z^{-1} + (mglT^2/I + 1 - DT/I)\, z^{-2}} \qquad (10)$$

where

$\tau(z)$ torque generated by actuator (input) ;

$\theta(z)$ angle of knee joint (output) ;

T sampling time (10msec) .

Parameters is represented by coefficient of following ARX model derived from equation(10)

$$\theta(k) + a_1\theta(k-1) + a_2\theta(k-2) = b_1\tau(k-1) + b_2\tau(k-2) + e(k) \quad (11)$$

$$I = \frac{T^2}{2b_1} \quad (12)$$

$$D = \frac{(a_1+2)T}{2b_1} \quad (13)$$

$$m = \frac{a_1+a_2+1}{2glb_1} \quad (14)$$

So parameters can be found by coefficient $a_1 \sim b_2$. If these parameters are unidentified, estimated coefficient vector and recursive vector are represented by

$$\hat{\alpha} = (a_1\ a_2\ b_1\ b_2) \quad (15)$$

$$\varphi(k) = [-\theta(k-1)\ \ -\theta(k-2)\ \tau(k-1)\ \tau(k-2)] \quad (16)$$

$$P(k) = \left(\sum_{t=1}^{k} \varphi(t)\varphi^T(t)\right)^{-1} \quad (17)$$

Then, estimated coefficient vector is found as follows

$$\hat{\alpha}(k) = \hat{\alpha}(k-1) + K(k)\left[\theta(k) - \varphi^T(k)\hat{\alpha}(k-1)\right] \quad (18)$$

$$K(k) = \frac{P(k-1)\varphi(k)}{\lambda + \varphi^T(k)P(k-1)\varphi(k)} \quad (19)$$

$$P(k) = \frac{1}{\lambda}\left[P(k-1) - \frac{P(k-1)\varphi(k)\varphi(k)^T P(k-1)}{\lambda + \varphi^T(t)P(t-1)\varphi(t)}\right] \quad (20)$$

where λ is forgetting factor ($0 < \lambda < 1$) . Fig. 5. shows experimental data of parameter estimation. The experimental subject (operator) is 25-years-old, physically unimpaired male(height: 176cm, weight: 68kg). The operator is seated putting on HAL-3 with weakness of muscles, and actuator of knee joint generate the torque randomly for irregular angle pattern. The length between joint and COG is determined as l=0.13[m][7], initial value of forgetting factor and recursive vector are selected as λ=0.998, $P(0) = \Delta I$, $\Delta = 12000$ in this experiment. Table 2 shows found parameters of both lower thigh. These parameters was confirmed as adequate value by [10] including exoskeleton.

C. Experimental Results for Simple Motion Assisted by Virtual Torque with Impedance Adjustment

Fig. 6. shows setup of experiment for simple motion. The operator makes his lower thigh move up and down as to set target angle and initial angle with assist of HAL-3 in this experiment (at 1sec interval) . Same motion has done without HAL-3 before this experiment

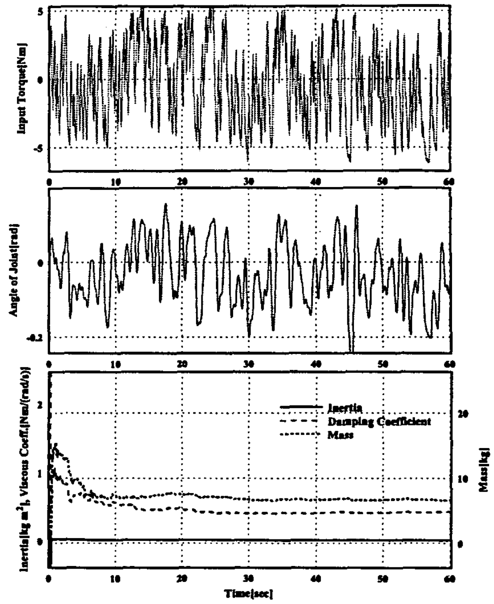

Fig. 5. Experimental result of parameter estimation (left lower thigh)

TABLE II
ESTIMATED PARAMETER OF OPERATOR'S LOWER THIGH (INCLUDING EXOSKELETON)

Parameter	Right Lower Thigh	Left Lower Thigh
Inertia[$kg \cdot m^2$]	0.0185	0.0179
Viscous Coef. [$N \cdot m/(rad/sec)$]	0.41	0.48
Mass [kg]	6.93	6.57

to compare the amplitude of s-EMG and calculate the Assist Efficiency. Fig. 7. indicate the experimental results. Fig. 7.-(a) represents the amplitude of filtered s-EMG and the angle of knee joint in the case of no motion aid by HAL, and 7.-(b) shows the case of motion assisted by HAL with only Virtual Torque. Though the power assist for same motion is performed according to set target angles, amplitude of s-EMG in Fig. 7.-(c) is remarkably reduced compared to Fig. 7.-(b), i.e., HAL-3 assists the torque of operator by 86% concerning the extensor of knee in the case of inertia-viscosity compensation, and operator can move the lower thigh with less power with intention to move. In Fig. 7., e.g., 30% compensation of inertia means reducing 30% of amount from the estimated inertia.

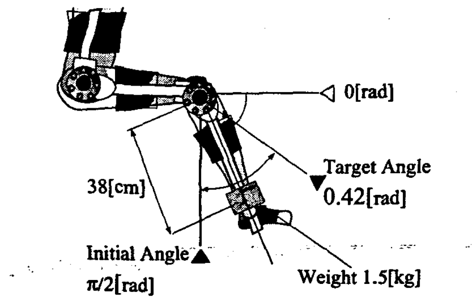

Fig. 6. Experimental setup for simple motion

TABLE III
ASSIST EFFICIENCY (AE) IN THE EXPERIMENT FOR SIMPLE ASSISTED MOTION

Assisting Method	AE_{fl}	AE_{ex}
Only Virtual Torque	62%	57%
+ inertia (30%) and viscosity compensation	83%	86%
+ inertia (60%) and viscosity compensation	86%	89%

D. Response Analysis of Operational Signal

Fig. 8. -(a), (b), (c) show the response properties of assisting methods. Shaded portions indicate the region where the angle of knee varies from initial angle to target. Operational signal of actuator rises gently at dotted circle in the case of only Virtual Torque (Fig. 8.-(a)) , and the time to peak is delayed for about 150msec compared to the raw signal of s-EMG due to filtering process. That causes a difference of response between human subject's leg and exoskeleton, and can make the operator feel discomfort for assisted motion although the actuator is controlled as to operator's intention. On the contrary, if inertia and viscous friction around knee joint are compensated, the change of operational signal becomes steep almost as soon as angle initiates falling, as Fig. 8.-(b), (c). Notably, the variation of operational signal precede that of the angle by 120 $\sim$ 130msec in Fig. 8. -(c). Hence, the operator can perform the intended motion of lower thigh including the exoskeleton smoothly by assist of HAL with less muscle force. Compensation ratio of inertia can be regulated as to operator's condition or external environment (e.g., grounding, not grounding) .

V. DISCUSSION

Static friction around knee joints are not compensated in this research because it is require to restrain the knee joint adequately by the exoskeleton when the actuator of each joint is in stationary state, e.g., in standing. Assist Efficiency subject to stay unchanged fairly from 70% of compensated inertia, and the motion of lower

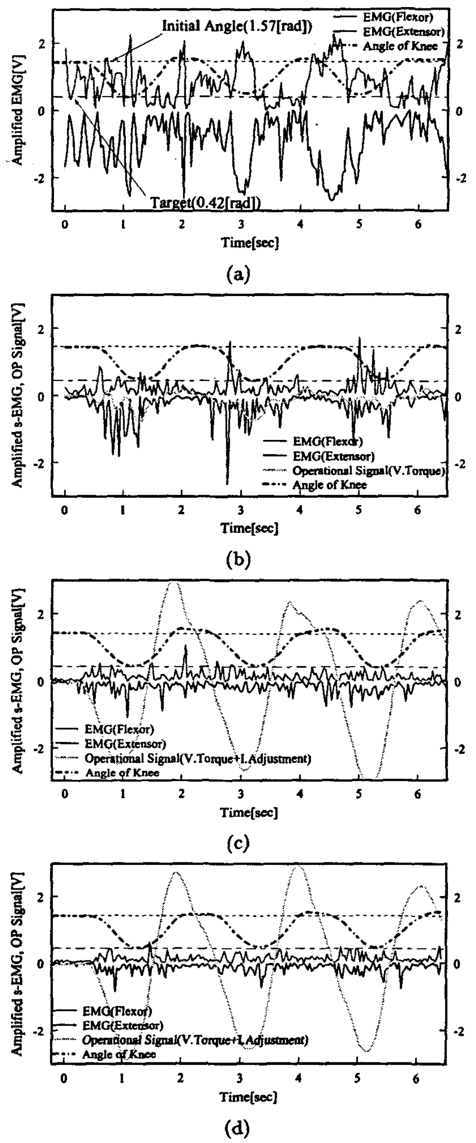

Fig. 7. Experimental result for simple motion (a)no assist with HAL, (b)Virtual Torque, (c)+ inertia (30%) and viscosity compensation, (d)+ inertia (60%) and viscosity compensation

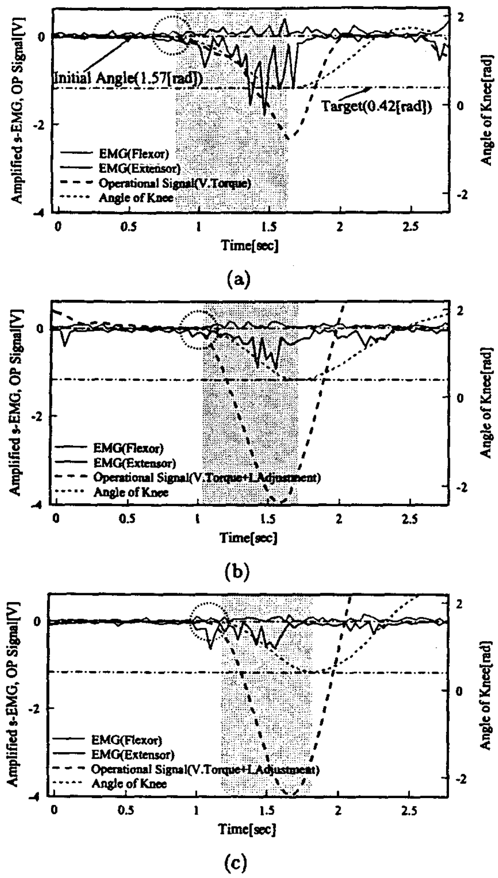

Fig. 8. The Response properties of assisting methods(a)Virtual Torque, (b)+ inertia (30%) and viscosity compensation, (c)+ inertia (60%) and viscosity compensation

thigh become unstable from approx. 95 % . Percentage of inertia to be compensated is found based on operator's condition, and also can be adjusted for operator to move the lower thigh lighter. 30% $\sim$ 50% is suitable by operator's opinion in this experiment.

VI. Conclusion

The paper has proposed the power assist control with HAL-3 using s-EMG and impedance adjustment around knee joint. We have developed the power assisting system, HAL-3 which can generate enough output to assist the motions at each joint, have a interactive performance by sensor system and wireless LAN. Virtual Torque based on s-EMG is adopted as basic control method of HAL-3, and we confirm the assisting motion as to intention of operator can be realized. For more effective power assist control, we suggest the impedance adjustment around knee joint. Parameters required for impedance adjustment like as inertia, viscous friction are estimated by Recursive Least Square method. The evaluation of assisted motion is carried out by Assist Efficiency calculated from average of s-EMG as its amplitude is almost in proportion to torque generated by muscle force. The experimental result for simple motion with impedance adjustment shows the operational signal of actuator is improved, and remarkable decrease of required torque at knee joint despite the same motion as only with Virtual Torque.

We will expand this method up to hip joint, and carry out the adaptive power assist control to the operator by auto-regulation of each parameter for whole joint of lower body as future works. And in near future, we plan to perform the experiment with the old or the disabled person who have functional disorder in the lower body.

Acknowledgments

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