Mechanism and Control of Knee Power Augmenting Device with Backdrivable Electro-Hydrostatic Actuator

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Abstract—Wearable robots that assist human mobility are actuated according to human intention. The estimation of the intention is difficult that the direct measurement is not possible. Especially the force control of the robot requires quick response and highly non-repeatable. Biological signals such as electromyography are often used but they lack stability. Sensitivity function gain maximization is another approach, but high backdrivability is necessary to realize comfort, which was difficult in conventional gear driven actuators. In this paper, we apply backdrivable electro-hydrostatic actuator to realize intrinsic backdrivability, which is then enhanced with low-impedance controller consisting of inertia scaling and disturbance observer based friction compensation. Power augmenting controller was realized with sensitivity function gain maximization. Numeric analysis and evaluation tests on actual prototype were carried out.

Keywords: power assist device, backdrivability, sensitivity function, electro-hydrostatic actuator

I. Introduction

Extending physical abilities of human are pursued in many ways. Usage of tools is one of the most common methods, however it requires skills of users to be effective. To improve the physical strength of human, power augmentation devices are studied since it is expected to expand the capability intuitively, requiring less skills for effective operation.

In late 1960’s, United States Military and General Electric Co. developed Hardiman prototype [1]. This project used master-slave architecture with valve controlled hydraulics to actuate whole body exoskeleton weighed 1500 lbs (680 kg). In recent years, control strategies that enable direct attachment of the robot output to the human body were realized, which contributed to the reduction of the weight and size of the exoskeleton.

In advanced countries, population ratio of elderly is increasing rapidly. In Japan, the population of the people aged above 65 is expected to grow above 40% of the total population by 2055 [2]. The impact on the society is extensive due to the increase load in nursery, growing social security budget, and decrease in working population. The power augmentation device is expected to play an important role in such society both from load reduction in nursery industry and independence of elderly point of view.

There are numerous power augmentation devices being studied [3], [4], [5], [6], [7]. Many of them [4], [6], [7] measure biological signals as EMG (Electromyography) or Muscle hardness to estimate the human intention. Use of EMG has an advantage that the intention can be estimated before the muscle tension is exerted since the myoelectric potential rises before the muscle activation. This phenomenon is known as electro-mechanical delay. However, the biological signals in general are unstable and noisy, making it difficult to use the amount of the signal without frequent calibration.

The method proposed by Kazerooni et al. [3] does not require measurement of human signal; no EMG nor no muscle force measurement. Instead, they increase the sensitivity of the joint controller to generate large joint velocity with small joint torque. This control scheme is interesting that the signal the controller requires is the encoder value, making sensing system reliable. However this method requires advanced backdrivability in joint actuation in order to eliminate discomfort upon wearing.

Previous studies [8], [9] have shown the efficacy of hydrostatic actuator for improvement of backdrivability. Hydrostatic actuation can reduce the friction in the transmis-
sion due to its reduction principle, and the internal leakage within the hydraulics improves the output backdrivability as stated in [9], [10]. This intrinsic backdrivability enables further enhancement of backdrivability with low-impedance controller.

The objectives of this paper is to propose the assisting control scheme of EHA driven knee power augmentation device targeted for elderly [11], [12]; combining low-impedance controller and sensitivity maximization. The assisting principle follows the method of [3], but the controller considers specific system structure to justify controller architecture. Discomfort of movement inconsistency between the user and the device can be minimized by backdrivability enhancement with low-impedance controller which include inertia scaling and friction compensation.

This paper is organized as follows. In section II, knee power augmentation device with EHA is presented. The EHA used is intrinsically backdrivable to enhance the comfort. In section III, model of EHA used throughout this paper is presented. In section IV, low-impedance control of the EHA is proposed. The controller consist of inertia scaling and observer based friction compensation. In section V, power augmentation controller based on sensitivity function maximization is proposed. In section VI, experiments to verify proposed control scheme is carried out. Section VII is the conclusions.

II. Knee Power Augmentation Device Using Electro-Hydrostatic Actuator

Our objective for this paper is to develop a single joint power augmentation device for knee joint. We focused on EHA as a actuator for power assist exoskeleton for following properties:
1. Force controllability. Since the actuator is backdrivable, we have more sensitivity and control over force.
2. Durability. In hydraulic system, force is transmitted with large surface area, which disturbs stress and enhances the durability.
3. Actuator placement freedom. In hydraulic system, there is placement freedom for pump placement. Also relative position between pump and hydraulic motor can be flexible. Heavy devices can be placed near the center of mass to enhance mobility.

Fig. 2 shows the outlook of the knee power augmentation device developed. The hydraulic schematic used in the device is shown in Fig. 3. Only hydraulic motor is attached to knee to reduce the weight of extremity (Fig. 4). Other portion as pump, motor, and battery are attached to hip. Pump and motor are shown in Fig. 5. Two passive joints with axes perpendicular to the hydraulic motor joint are placed in the attachment linkages as in Fig. 6 [12] to resolve axis misalignment induced by screw-home movement [13].

Output power rating was chosen using statistics of Japanese elderly to provide enough torque and speed for daily life [12]. The specification of the device is shown in Table I.

III. Model of Electro-Hydrostatic Actuator

EHAs are a class of servo motor driven displacement control type hydraulic system with typical architecture shown in Fig. 3. The dynamics of an EHA is given as a set of following equations.

\[
J_i\ddot{\theta}_i = -k_{i3}p_i - \tau_{if}(\dot{\theta}_i, p_i, \dot{p}) + \tau_i = -k_{i1}k_{i3}\dot{\theta}_i + k_{i2}k_{i3}\dot{\theta}_i - \tau_{if}(\dot{\theta}_i, \dot{\theta}_i, \dot{p}) + \tau_i
\]

\[
(i, \ddot{i}) \in \{(1, 2), (2, 1)\}
\]

<table>
<thead>
<tr>
<th>Description</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Oil Viscosity</td>
<td>100cSt</td>
</tr>
<tr>
<td>Total Reduction Ratio</td>
<td>1:449</td>
</tr>
<tr>
<td>Maximum Torque</td>
<td>30 Nm at 10RPM</td>
</tr>
<tr>
<td>Maximum Speed</td>
<td>55 RPM at 0Nm</td>
</tr>
<tr>
<td>Total Weight</td>
<td>1124 g</td>
</tr>
</tbody>
</table>
Here, the parameters used in the equation is explained in Table II. $k_{ij}$ are constants determined by form and hydraulic properties, such as pump thickness and hydraulic viscosity. Subscript $i$ are either 1 or 2; 1 show the primary (input) side, thus pump parameters and 2 show secondary (output) side, thus hydraulic motor parameters. Subscript $i$ and superscript $i$ shows other side of $i$, thus if $i = 1$ then $i = 2$ and Vice Versa.

Relation between pressure difference $p_i$ and pump/hydraulic motor speed is explained as (3).

$$p_i = k_{1i} \dot{\theta}_i - k_{2i} \dot{\theta}_i$$

Table II. Nomenclature

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description (units)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\theta_i$</td>
<td>Position m</td>
</tr>
<tr>
<td>$J_i$</td>
<td>Mass kg</td>
</tr>
<tr>
<td>$\tau_{if}$</td>
<td>Friction force N</td>
</tr>
<tr>
<td>$\tau_i$</td>
<td>Input force N</td>
</tr>
<tr>
<td>$p_i$</td>
<td>Pressure difference Pa</td>
</tr>
<tr>
<td>$\dot{p}$</td>
<td>Amount of pressurization Pa</td>
</tr>
</tbody>
</table>

IV. Low-Impedance Controller for Electro-Hydrostatic Actuators

The EHA we use in designed to have low mechanical friction and backdrives with the torque less than 0.2 Nm [12]. This is below 0.7% of the rated maximum torque and it is a small value. However, for farther enhancement of backdrivability, control should be introduced.

The apparent inertia of the actuator and frictions are two major caused of backdrivability degradation. To reduce these effects following controllers are effective.

1. Inertia scaling: Reduces apparent inertia of the actuator by torque feedback.
2. Friction compensation: Cancels friction torque with estimated friction torque.

In this paper, we combine inertia scaling using torque estimated from pressure difference, and friction compensation based on momentum based friction observer are used to achieve low mechanical impedance and farther enhancement of backdrivability.

A. Inertia Scaling

Inertia scaling is the control technique to reduce apparent inertia of the actuator with torque feedback control. For the actuator with the primary side inertia $J$ and reduction ratio $N$, the reflected inertia from secondary side becomes $JN^2$ which can become significantly large with large $N$ even $J$ is small. In modern actuators, high speed - high power motors are often used. This mean the usage of high reduction ratio is innate, which lead to large reflected inertia.

In our system, the total reduction ration of the system is 1:449. Hence it is crucial for the backdrivability enhancement to reduce primary side inertia.

In torque sensor based systems, Ott et al. [14] used impedance controller with inertia scaling on the primary side to realize passive controller with low apparent inertia. We have used similar controller on EHA [9]. In hydraulic actuators, the exerted torque can be estimated with pressure
Inertia scaling control was done by pressure feedback as in (4) to the system (1).

\[
\tau_1 = \frac{1}{\alpha}u + \left(1 - \frac{1}{\alpha}\right)k_{13}p_1 \tag{4}
\]

\[
\alpha = \frac{\hat{J}_1}{J_1} \tag{5}
\]

Here, \(\hat{J}_1\) is the desired pump inertia, usually chosen smaller than the physical value to enhance backdrivability. \(k_{13}\) is the estimated value of \(k_{13}\), and \(u\) is the intermediate control input, which is used instead of \(\tau_1\). Considering the purpose of reducing the inertia, \(\alpha\) is chosen as \(0 < \alpha < 1\). See Fig. 7 for the controller structure.

It is obvious that the pressure feedback (4) modifies the dynamics of the system to (6) when \(k_{13} = k_{13}^*\) holds.

\[
\alpha J_1 \dot{\theta}_1 = -k_{11}k_{13}\dot{\theta}_1 + k_{12}k_{13}\dot{\theta}_2 - \tilde{\tau}_{1f} + u \tag{6}
\]

Here, \(\tilde{\tau}_{1f} = \alpha\tau_{1f}\). Thus pump friction is reduced by this feedback as well. The stability of the feedback (4) is discussed in [10].

Since the inertia scaling does not change dynamics structure, hereafter \(\tau_1\) and \(\alpha\) are used without distinction. When the inertia scaling is used, \(\tau_1\) means the intermediate input \(u\).

B. Friction Compensation Friction Observer

Although the inertia scaling partly compensates friction at pump, it is still desirable to compensate pump side friction torque. Also, from previous investigation, it was reported that the unmodeled friction at pump oil seal affects precision performance of output torque control [12]. Hence, it is important to compensate as much friction torque as possible to maximize comfort of wearing.

Many friction compensation methods require friction model [15], [16], [17]. This however requires naive treatment of friction parameters since the friction torque is greatly affected by operating conditions as temperature, running time, and operating pressure.

In this research, we use disturbance observer for friction estimation, which then is used as feed-forward term for compensation as in Tien et al. [18]. This method is momentum based disturbance observer proposed by De Luca and Mattone [19], [20] that does not require acceleration measurement. We have made modifications on this method to apply it to the hydraulic system.

The friction torque \(\tau_{1f}\) is estimated as \(\hat{\tau}_{1f}\), with the disturbance observer given with (7), (8). Estimated torque \(\hat{\tau}_{1f}\) is feed-forwarded to compensate the friction torque.

\[
\tau_1 = J_1\ddot{\theta}_1 + k_{13}p_1 + \tilde{\tau}_{1f} \tag{7}
\]

\[
\hat{\tau}_{1f} = -LJ_1(\dot{\hat{\theta}}_1 - \hat{\theta}_1) \tag{8}
\]

Here, \(L\) is the observer gain and \(\hat{\theta}_1\) is the expected angular speed when the torque is applied to the nomina model (the model without friction). The estimated friction \(\hat{\tau}_{1f}\) converges to the pump friction \(\tau_{1f}\).

As the implementation, (9) is used instead of (7).

\[
\tau_1 = J_1\ddot{\theta}_1 + k_{13}p_1 + \tilde{\tau}_{1f} \tag{9}
\]

In (9), the term of \(k_{13}p_1\) is used instead of \(k_{13}\) to include the pressure loss due to fluid friction in the connecting tubes between hydraulic motor unit and the power unit, into the friction compensation term \(\tilde{\tau}_{1f}\). Also, in the controller, identified value of \(J_1^*\) used for \(J_1\).

The block diagram of friction compensation based on disturbance observer used in this research is shown in Fig. 8. In implementation, the torque command to the pump is realized by vector current control of brushless DC motor [21] and measured motor torque constant \(K_T\).

V. Power Augmentation Control for Electro-Hydrostatic Actuators

To simplify the notation, the dynamics of an EHA (1) can be re-written as (10).

\[
\begin{align*}
\dot{\theta}_1 &= d_{11}\dot{\theta}_1 + d_{12}\dot{\theta}_2 - \tau_{1f}/J_1 + \tau_1/J_1 \\
\dot{\theta}_2 &= d_{22}\dot{\theta}_2 + d_{21}\dot{\theta}_1 - \tau_{2f}/J_2 + \tau_2/J_2
\end{align*}
\tag{10}
\]

By choosing \(\tau_1\) as a control input, \(\dot{\theta}_2\) as system output, and \(\tau_2\) as disturbance input, (10) can be expressed as a state-space model as follows:

\[
\dot{x} = Dx - T_f + B_1\tau_1 + B_2\tau_2 \tag{11}
\]

\[
\dot{\theta}_2 = Cx \tag{12}
\]

\[
x = \begin{bmatrix} \dot{\theta}_1 & \dot{\theta}_2 \end{bmatrix}^T \tag{13}
\]

\[
D = [d_{ij}] \in \mathbb{R}^{2 \times 2}
\]

\[
T_f = \begin{bmatrix} \tau_{1f}/J_1 & \tau_{2f}/J_2 \end{bmatrix}^T
\]

\[
B_1 = \begin{bmatrix} 1/J_1 & 0 \end{bmatrix}^T, B_2 = \begin{bmatrix} 0 & 1/J_2 \end{bmatrix}^T
\]

\[
C = \begin{bmatrix} 0 & 1 \end{bmatrix}
\]

The sensitivity function \(S\) is the transfer function from the disturbance input to the system output. Neglecting the
When the system has the structure of Fig. 1, total state feedback gain becomes $B_1H_1 + B_2\theta h_{22}$. $H_1$ must be chosen within the range that the total system becomes stable.

In the system with this feedback, sensitivity function is given as follow using $D' = D + B_1H_1 + B_2\theta h_{22} = [d_1']$.

$$S = C(sI - D')^{-1}B_2 = \frac{1}{J_2} \frac{s - d_{11}'}{s^2 - (d_{11}' + d_{22})s + (d_{11}'d_{22} - d_{12}'d_{21})}$$  \hspace{1cm} (18)

The gain of this function is calculated as (19).

$$|S(j\omega)| = \frac{1}{J_2} \frac{|j\omega - d_{11}'|}{|d_{11}'d_{22} - d_{12}'d_{21} - \omega^2| - j(d_{11}' + d_{22})\omega}$$  \hspace{1cm} (19)

Considering the form of (19) and the fact that $h_{22}$ is constant, it is expected so the increment of $h_{11}, h_{12}$ under the condition that the system becomes stable increases the amount of $|S|$. Since the human controller $h_{22}$ is the stabilizing controller, we can assume it being $h_{22} < 0$. Hence, it is expected that feeding back $\dot{\theta}_2$ with positive gain to $\tau_1$ increases $|S|$. This stability region is related with the human controller parameter as mentioned in [22]. This positive state feedback makes the actuator control loop unstable, but together with human musculoskeletal controller $h_{22}$, the system maintains the stability.

To verify this idea, numerical analysis was performed with the parameters of developed EHA. The parameters were estimated from CAD data that are shown in Table III. We assumed that the inertia ratio in CAD data is 1/3 of the value in Table III. We did not include the inertia of human limb in the secondary side inertia, but together with human musculoskeletal controller $h_{22}$, the system maintains the stability.

There is some stability margin for the value of $h_{11}$ in positive region, but this amount is very small. Hence we put $h_{11} = 0$ for this analysis. Feedback gain for human controller was fixed to $h_{22} = -100$. There is no clear reasoning for the amount of this value, but the value was chosen to make the system stable and the amount that is expected to be feasible. Amount of $h_{22}$ changes the amount of $h_{12}$ that make the system stable, but it does not affect the trend of analysis with regard to the change of $h_{12}$.

The bode plot of the $S$ with variable amount of $h_{12}$ swept in the range of $-1.2 \leq h_{12} \leq 1.2$ is shown in Fig. 9. Fig. 9 shows that the system maintains stability to some positive value of $h_{12}$, but turns to instability with larger amount of $h_{12} > 0$. However, there is trend that the gain of sensitivity function $S$ increases as the system reaches the stability limit. From this result, it was shown that the positive feedback gain of $h_{22}$ in stability region increases the gain of sensitivity function $S$. This implies that the such controller acts as power augmenting controller as stated above.

This simplification is reasonable because viscosity has the stabilization effect on the system and Coulomb friction in the designed system is sufficiently small.
TABLE III. Dynamics Parameter of EHA Obtained from CAD Design

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$d_{11}$</td>
<td>$-0.0603 \times 10^4$ 1/s</td>
</tr>
<tr>
<td>$d_{12}$</td>
<td>$7.94 \times 10^4$ 1/s</td>
</tr>
<tr>
<td>$d_{21}$</td>
<td>$0.0624 \times 10^4$ 1/s</td>
</tr>
<tr>
<td>$d_{22}$</td>
<td>$-8.22 \times 10^4$ 1/s</td>
</tr>
<tr>
<td>$J_1$</td>
<td>$2.38 \times 10^{-3}$ kgm^2</td>
</tr>
<tr>
<td>$J_2$</td>
<td>$8.83 \times 10^{-4}$ kgm^2</td>
</tr>
</tbody>
</table>

However, it should be noted that the analysis above assumes rigid coupling between human and the exoskeleton. In reality there are compliances and additional masses in the coupling that complicate the dynamic behavior of the system with uncertain unstable parameters. For this reason, normally gain margin of the power augmenting controller must be chosen with some large value to suppress the vibration due to the unmodeled dynamics of the coupling. Furthermore, it is difficult to estimate what the human controller parameter is; most certainly the feedback controller of human is time varying and nonlinear. Thus there is only small meaning of trying to make quantitative discussion of maximum amount of stable $h_{12}$ with identified $h_{22}$. Hence it is reasonable to make $h_{12}$ a tuning parameter that is calibrated on actual hardware upon the usage.

VI. Experiments

A. Evaluation of Low-Impedance Controller

In order to validate the efficacy of the disturbance observer based friction compensation, a step response of the EHA with compliance control was tested. The difference in remaining friction is observed from the difference in the response trajectory.

In this test, the EHA was fixed to ground with configuration that the joint axis is parallel to the ground. Inertia scaling of 1:5 was applied to the system and the compliance of 4 Nm/rad was realized with control. The step input was applied by releasing load torque. Initial deflection was given to the system to apply initial load. At the time 0, the link was released so the link returns to the resting angle which is set to the vertical posture.

Fig. 10 shows the step response of the controller with and without friction compensation. Initial deflection of approximately 0.8 rad, thus approximately 3.2 Nm of initial load was applied to the joint. The load was released at time 0.5 sec. The response of the joint was over damping for the case without friction compensation, where it changed to damping oscillation for the case with friction compensation. This change in oscillation mode shows the significant reduction of friction torque.

From the result, efficacy of the low-impedance controller of combining inertia scaling and disturbance observer based friction compensation was experimentally proven.

B. Power Augmentation Controller

In this paper, power augmentation control is done by maximization of sensitivity function gain $|S|$. As explained in section V, this is done by positive feedback of output joint velocity $\dot{\theta}_2$ to primary pump torque $\tau_1$ in stable region.

The evaluation is done by comparing measured operator torque for same load under various gain configuration. The
TABLE IV. Wire Routing Configuration Parameters (Units are in mm)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Load Wire</th>
<th>Operator Wire</th>
</tr>
</thead>
<tbody>
<tr>
<td>$l$</td>
<td>310</td>
<td></td>
</tr>
<tr>
<td>$h$</td>
<td>-277</td>
<td>277</td>
</tr>
<tr>
<td>$v$</td>
<td>200</td>
<td>310</td>
</tr>
</tbody>
</table>

Load torque and operator torque are applied through wires as in Fig. 11. The wires were then routed using pulley as in Fig. 12 with parameters in Table IV. For the load torque wire, a weight was attached to the wire to apply tension. The operator torque was calculated from the wire tension measured with force gauge. Inertia scaling of 1:8 was applied for the results with friction compensation. The operator torque was applied from the relaxed position (approximately $\theta_2 = -0.7$ rad) to the upright position ($\theta_2 = 0$), against the load torque.

The result is shown in Fig. 13. The figure shows the difference in operator torque with regard to the positive feedback gain. Joint angle trajectories of all test patterns are similar and the load torque trajectories are overlapping, but there is clear distinction between operator torque trajectories.

In all cases, the load starts moving when the operator torque exceeds the load torque. In the test case only with friction compensation, the torque trajectory after the movement of joint roughly coincides. The amount above the load torque is the addition of measurement error and residual of friction compensation. However, this still is a significant reduction from the case with no control shown with two dotted dashed line. The operator torque reduces as the $h_{12}$ increases, that was estimated behavior.

To see this result from energy point of view, the result calculated in energy gain and consumption is shown in Fig. 14. In all cases, the load gained about 207 J, with almost same time trajectory (upper plot). However, what operator has consumed in the case with low-impedance control was 211 J, where the result with maximum gain in the plot was 153 J, which is 27% reduction of energy consumption, or in other words, 27% augmentation of the power was confirmed. As a reference, the result of operator consumed energy without control was 288 J.

As mentioned in section V, gravity compensation must be used in together with proposed controller to further reduce the operating torque because current controller does not compensate static torque.

C. Power Augmentation Control Evaluation with Human Muscle Activity Analysis

To see the efficacy of the method when the device is attached to the knee, standing up from a chair was performed with and without power augmenting device under assist control. The muscle activation was estimated using iEMG (integrated electromyography). EMG was measured with sEMG (surface EMG) of Vastus lateralis. Amount of iEMG was normalized with EMG at MVC (maximum voluntary contraction).

Fig. 15 shows the measurement result of joint angle trajectory and iEMG. The joint angle is only measured for the case with the power augmenting device, but the data of the EMG is synchronized to the reference joint trajectory. From the observation of EMG, the average amount of iEMG was reduced by 30% with the presence of the power augmenting device, which show the efficacy of the device. Same as in previous section, farther reduction of muscle activity is expected when the controller is combined with gravity compensation.

The damping oscillation observed after the knee extension is due to the unmodeled dynamics of human-exoskeleton coupling. The suppression of this oscillation is still an open problem.

VII. Conclusions

The importance of lower limb exoskeleton for power augmentation is rising. For these devices to be useful, stable control method must be investigated.

In order to realize stable power augmentation, control strategy not relying on biological signal is desirable. Sen-
Fig. 13. Torque and Joint Position Profile of Power Augmenting Control Evaluation. Gain Settings: blue dotted=0, red dashed=5, green dotted dashed=10, black solid=15. Black dashed line with two dots show the result with no control.

In this paper, we proposed the power augmentation control method that combine low-impedance controller and sensitivity function gain maximization. Followings are the conclusions obtained in this paper.

1. A low-impedance controller for electro-hydrostatic actuators was proposed. This method combines inertia scaling and disturbance observer based friction compensation. The proposed friction observer uses pressure sensors instead of torque sensors, which was used in original method. The controller compensates not only pump friction, but also the pressure loss due to the fluid friction in the transmission tubes.

2. The efficacy of the low-impedance controller was confirmed on actual EHA for the knee power augmentation device with step response. The oscillation mode of the response changed due to the significant reduction of the friction.

3. A power augmenting control based on sensitivity function gain maximization was proposed. The method proposed by Kazerooni et al. was modified to reflect the dynamics of EHA. The controller feeds back secondary side angular velocity to primary side torque command with positive gain. The effect of the controller was numerically analyzed.

4. The power augmenting controller was tested on a test rig. The energy consumption of operator was reduced by 25% in the test performed. The controller was also tested on human. The evaluation was done with iEMG (integrated electromyography) of Vastus lateralis normalized with iEMG at maximum voluntary contraction. 30% reduction of iEMG was observed with the presence of power augmentation device.

Acknowledgments

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References


Fig. 14. Energy Profile of Load and Operator of Power Augmenting Control Evaluation. Gain Settings: blue dotted=0, red dashed=5, green dotted dashed=10, black solid=15. Black dashed line with two dots show the result with no control.


Fig. 15. Muscle Activity Comparison on Presence of Power Augmenting Device. Green dashed line shows the reference joint trajectory for human. Red dotted line shows the trajectory of the case with power augmenting device. Black solid line shows the case without power augmenting device.