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Motion Control of Wearable Walking Support System with Accelerometer Considering Swing Phase Support

Yasuhisa Hirata, Takuya Iwano, Masaya Tajika and Kazuhiro Kosuge

***Abstract*— In this paper, we propose a method for supporting a walking of a human by using wearable walking support system called Wearable Walking Helper. In this method, the support moment for the knee joint of the user is calculated by the human model and a part of it is supported by the actuator of the Wearable Walking Helper. In the conventional method we proposed so far, we only consider the stance phase of the gait which requires a large moment of the knee joint in walking for supporting the weight of the user. In this paper, we especially focus on the support of the swing phase of the gait which requires a moment of knee joint for supporting not only the weight of the leg but also the weight of the Wearable Walking Helper itself. For realizing the swing phase support, we measure an inclination of the upper body of the user with respect to the vertical direction by using the accelerometer and apply it to the calculation method using human model for deriving the support moment of the knee joint. The proposed method is applied to the Wearable Walking Helper experimentally and the experimental results illustrate the validity of the proposed method.**

I. Introduction

Due to the rapid aging of the population in recent years, the number of people in need of care is increasing. With the development of the robot technologies, various kinds of rehabilitation system or human assist robot such as walking aid system and manipulation aid system have been developed for supporting the handicapped people including the elderly. Several wearable assist systems have also been developed [1–5] for supporting the daily activities of the people such as walking, handling and so on. In this paper, we especially focus on a wearable walking support system for the purpose of assisting people who have difficulties in walking because of weakened leg muscles to walk.

Most of conventional wearable human assist systems proposed so far try to identify the motion patterns of the user based on the biological signals such as EMG (electromyogram) signals or hardness of skin surface, and they are controlled to assist the user based on the identified motions. However, the noises included in these biological signals make it difficult to identify the motions of the user accurately. In addition, since each joint of a human body is actuated with the cooperation of many muscles, it is difficult to identify the motions of the user based on the activities of only few muscles observed by EMG signals or hardness of skin surface.

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To overcome these problems, we developed a wearable walking support system which is able to support walking activity without using biological signals as shown in Fig.1. In our system, the support moment of the joints of the user is calculated by an approximated human model of four-link open chain mechanism on the sagittal plane and a part of the joint moment is assisted by the actuator of the wearable walking support system [6].

In the conventional control algorithm, we assumed that we only support the stance phase of the gait which requires a large moment of knee joint for supporting the weight of the user, and the weight of the support device is neglected in the stance phase. We also assumed that the user stands on flat ground and inclination of Foot Link is always parallel to the ground. However, if we consider the support of the swing phase of the gait in which we could not neglect the weight of the support device, the conventional control algorithm makes the burden of the user increase, because the user has to lift the support device in the swing phase. Additionally, the inclination of Foot Link always changes widely in the swing phase.

In this paper, to enable swing phase support, we derive the support moment for the knee joint to guarantee the weight of the device, and then we propose a method for measuring the inclination of a link of the human model with respect to the vertical direction by using an accelerometer and apply it to the calculation method using human model for deriving the support moment of the joints in both stance phase and swing phase. The proposed method is applied to the Wearable Walking Helper experimentally and the experimental results illustrate the validity of the proposed method.

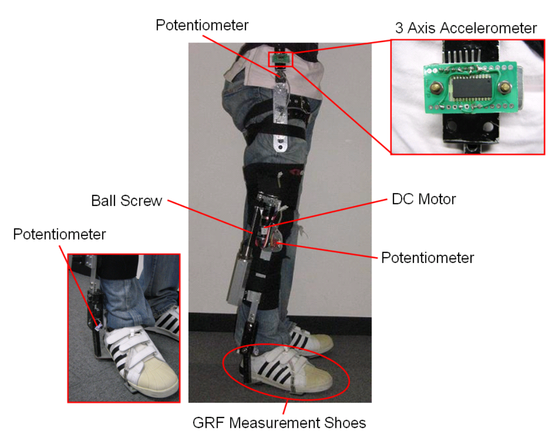


Fig. 1. Wearable Walking Helper with Accelerometer

II. Wearable Walking Helper

In this section, we briefly introduce a developed wearable walking support system referred to as Wearable Walking Helper. We developed the smaller and lighter support device for the knee joint than its conventional system proposed in [6]. Fig.1 shows the prototype of the system which consists of knee orthosis, prismatic actuator and sensors. The knee joint of the orthosis has one degree of freedom rotating around the center of the knee joint of the user on sagittal plane. The mechanism of the knee joint is a geared dual hinge joint. The prismatic actuator, which is manually back-drivable, consists of DC motor and ball screw. By translating the thrust force generated by the prismatic actuator to the frames of the knee orthosis, the device can generate support moment around the knee joint of the user.

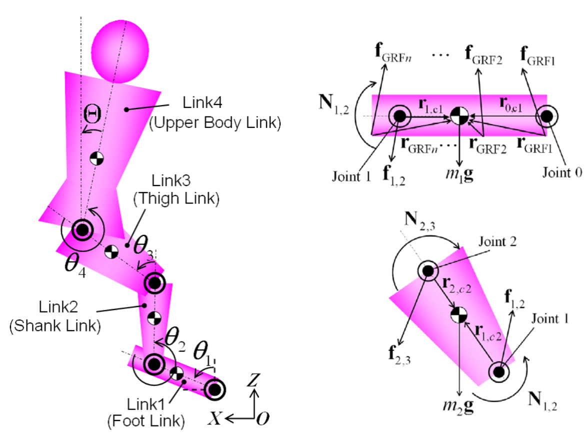
Three potentiometers are attached to the ankle, knee and hip joints to measure the rotation angle of each joint. To measure the Ground Reaction Force (GRF), two force sensors are attached to the shoe sole: one is on the toe and the other is on the heel. In addition, to measure inclination of the link, a 3-axis accelerometer is attached to near the hip joint. By using measured joint angles, GRFs and the inclination of the link, the support moment for the knee joint is calculated based on the human linkage model.

III. Model-Based Control Algorithm

In this section, we describe the control algorithm of the wearable walking support system. Firstly, we derive the knee joint moment based on an approximated human model. Secondly, we propose the method for deriving the knee joint moment to support the weight of the device itself. Then the support joint moment, which should be generated by the actuator of the support device, is calculated.

A. Calculation of Knee Joint Moment Using Human Model

To control Wearable Walking Helper, we use an approximated human model as shown in Fig.2. Under the assumption that the human gait is approximated by the motion on the sagittal plane, we consider only $Z-X$ plane. The human model consists of four links, that is, Foot Link, Shank Link, Thigh Link and Upper Body Link and these links compose a four-link open chain mechanism.



(a) Coordinate Systems (b) Foot and Shank Links

Fig. 2. Human Model

To derive joint moments, we first set up Newton-Euler equations of each link. At the link i, Newton-Euler equations are derived as follows;

$$\mathbf{f}_{i-1,i} - \mathbf{f}_{i,i+1} - m_i\mathbf{g} = m_i\dot{\mathbf{v}}_{\mathrm{c}_i} \tag{1}$$

$$N_{i-1,i} - N_{i,i+1} + \mathbf{r}_{i,c_i} \times \mathbf{f}_{i,i+1} - \mathbf{r}_{i-1,c_i} \times \mathbf{f}_{i-1,i} = I_i\frac{d\dot{\theta}_i}{dt} \tag{2}$$

where, $\mathbf{f}_{i-1,i}$ and $\mathbf{f}_{i,i+1}$ are reaction forces applying to the joint i and $i+1$ respectively. m_i is the mass of the link i, $\mathbf{g}$ is the vector of gravity acceleration and $\dot{\mathbf{v}}_{\mathrm{c}_i}$ is the translational acceleration of the gravity center of the link i. $N_{i-1,i}$ and $N_{i,i+1}$ are the joint moments applying to the joint i and $i+1$ respectively. r_{i,c_i} is the position vector from the joint i to the gravity center of the link i and r_{i-1,c_i} is the position vector from the joint $i-1$ to the gravity center of the link i. I_i is the inertia of the link i and θ_i is the rotation angle of the joint i.

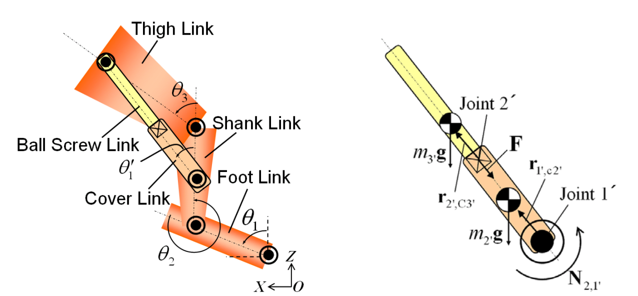
The knee joint moment $N_{2,3}$ can be derived by using the equations of foot link and shank link as follows;

$$\begin{aligned}\tau_{\mathrm{k}} = N_{2,3} = &- I_1\frac{d\dot{\theta}_1}{dt} - I_2\frac{d\dot{\theta}_2}{dt}\\ &- m_1(\mathbf{r}_{1,\mathrm{c}_1} - \mathbf{r}_{1,\mathrm{c}_2} + \mathbf{r}_{2,\mathrm{c}_2}) \times (\dot{\mathbf{v}}_{\mathrm{c}_1} - \mathbf{g})\\ &- m_2\mathbf{r}_{2,\mathrm{c}_2} \times (\dot{\mathbf{v}}_{\mathrm{c}_2} - \mathbf{g})\\ &+ (\mathbf{r}_{1,\mathrm{c}_1} - \mathbf{r}_{1,\mathrm{c}_2} + \mathbf{r}_{2,\mathrm{c}_2}) \times \sum \mathbf{f}_{\mathrm{GRF}}\\ &- \sum(\mathbf{r}_{\mathrm{GRF}} \times \mathbf{f}_{\mathrm{GRF}})\end{aligned} \tag{3}$$

where $\mathbf{f}_{\mathrm{GRF}}$ is Ground Reaction Force exerted on the foot link.

B. Calculation of Knee Joint Moment Considering Device Model

To derive the knee joint moment affected by the weight of the walking support system, we use the device model as shown in Fig.3. Since the device model has a closed-loop mechanism, it is impossible to easily derive the joint moment. A variety of schemes for deriving joint torques for robots consisting of closed chain mechanisms have been



(a) Device Coordinate Systems (b) Ball Screw and Cover Link

Fig. 3. Device Model

proposed [7–9]. In this paper, we apply the method proposed by Luh et al. [7].

First, we define that joint $1'$ is the connecting point between the Cover Link and Shank Link, joint $2'$ is position of the prismatic joint of the support device as shown in Fig.3(b) and joint $3'$ is the connecting point between the Thigh Link and Ball Screw Link. The closed-chain is virtually cut open at the joint $3'$ and we analyze it as virtual open-chain mechanism.

Next, the holonomic constraints are applied to the virtually cut joint. As a result, we can consider the spatial closed-chain linkage as a tree-structured open-chain mechanism with kinematic constraints. Similarly to the method we derived the knee joint moment using human model, the joint moments $N'_{2,3}$ which expresses the joint moment around joint 2 considering the effect of the support device and $N_{2,1'}$ can be derived based on Newton-Euler formulation as follows;

$$\begin{aligned} N'_{2,3} = &- I'_1 \frac{d\dot{\theta}_1}{dt} - I'_2 \frac{d\dot{\theta}_2}{dt} - I_{2'} \frac{d\dot{\theta}_{2'}}{dt} \\ &- m'_1(\mathbf{r}'_{1,c_1} - \mathbf{r}'_{1,c_2} + \mathbf{r}'_{2,c_2}) \times (\dot{\mathbf{v}}'_{c_1} - \mathbf{g}) \\ &- m'_2 \mathbf{r}'_{2,c_2} \times (\dot{\mathbf{v}}'_{c_2} - \mathbf{g}) \\ &- m_{2'} \mathbf{r}_{2',c_2} \times (\dot{\mathbf{v}}_{c_{2'}} - \mathbf{g}) \\ &- m_{3'} \mathbf{r}_{3',c_2} \times (\dot{\mathbf{v}}_{c_{3'}} - \mathbf{g}) \end{aligned} \tag{4}$$

$$\begin{aligned} N_{2,1'} = I_{2'} \frac{d\dot{\theta}_{2'}}{dt} &- m_{2'} \mathbf{r}_{1',c_{2'}} \times (\dot{\mathbf{v}}_{c_{2'}} - \mathbf{g}) \\ &- m_{3'} \mathbf{r}_{1',c_{2'}} \times (\dot{\mathbf{v}}_{c_{3'}} - \mathbf{g}) \end{aligned} \tag{5}$$

From the Newton equation of Cover Link, the generalized force $\mathbf{F}$ shown in Fig.3(b) can be derived as follows:

$$\mathbf{F} = m_{3'}(\dot{\mathbf{v}}_{c_{3'}} - \mathbf{g}) \tag{6}$$

where $\mathbf{F} = [F_{x2'} \ F_{z2'}]^T$ is two dimensional vector of generalized force and $F_{x2'}$ is zero since we only consider the gravity direction (z-axis) effected by the weight of the support device.

Now we consider the holonomic constraints for the virtually cut joint. The homogeneous transformation matrix from the joint $1'$ to the joint $3'$ through the joint 2 is

$$\begin{aligned} \mathbf{A}^2_{1'} \mathbf{A}^{3'}_2 &= \begin{bmatrix} \cos\theta_2 & 0 & \sin\theta_2 & l'_3 \sin\theta_2 \\ 0 & 1 & 0 & 0 \\ -\sin\theta_2 & 0 & \cos\theta_2 & l'_2 + l'_3\cos\theta_2 \\ 0 & 0 & 0 & 1 \end{bmatrix} \\ &= \begin{bmatrix} \mathbf{R}^{23'}_{1'2} & \mathbf{P}^{23'}_{1'2} \\ 0 & 1 \end{bmatrix} \end{aligned} \tag{7}$$

and similarly from the joint $1'$ to the joint $3'$ through the joint $2'$ is

$$\begin{aligned} \mathbf{A}^{2'}_{1'} \mathbf{A}^{3'}_{2'} &= \begin{bmatrix} \cos\theta'_1 & 0 & \sin\theta'_1 & d\sin\theta'_1 + l_b \sin\theta'_1 \\ 0 & 1 & 0 & 0 \\ -\sin\theta'_1 & 0 & \cos\theta'_1 & d\cos\theta'_1 + l_b\cos\theta'_1 \\ 0 & 0 & 0 & 1 \end{bmatrix} \\ &= \begin{bmatrix} \mathbf{R}^{2'3'}_{1'2'} & \mathbf{P}^{2'3'}_{1'2'} \\ 0 & 1 \end{bmatrix} \end{aligned} \tag{8}$$

The support device has a closed chain mechanism, and Thigh Link and Ball Screw Link are actually connected at the joint $3'$. Therefore, position vectors shown in equations (7) and (8) satisfy the following constraints.

$$\begin{aligned} \mathbf{c} &= \mathbf{P}^{23'}_{1'2} - \mathbf{P}^{2'3'}_{1'2'} \\ &= \begin{bmatrix} l'_3 \sin\theta_2 - (d + l_b)\sin\theta_1 \\ l'_2 + l'_3\cos\theta_2 - (d + l_b)\cos\theta'_1 \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \end{bmatrix} \end{aligned} \tag{9}$$

By using the generalized force and moments vector $\tau^o = [N'_{1,2} \ N'_{2,3} \ F_{z2'} \ N_{2,1'}]^T$ and considering the holonomic constraints, the following equation is satisfied;

$$\mathbf{J}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{f}(\ddot{\mathbf{q}}, \mathbf{q}) + \mathbf{g}(\mathbf{q}) - \tau^o + \left(\frac{\partial \mathbf{c}}{\partial \mathbf{q}}\right)^T \lambda = \mathbf{0} \tag{10}$$

where

$$\left(\frac{\partial \mathbf{c}}{\partial \mathbf{q}}\right)^T = \begin{bmatrix} 0 & 0 \\ l'_3\cos\theta_2 & -l'_3\sin\theta_2 \\ -\sin\theta_1\prime & -\cos\theta'_1 \\ -(d + l_b)\cos\theta'_1 & (d + l_b)\sin\theta'_1 \end{bmatrix} \tag{11}$$

Additionally, in the equation (10), the inertia term $\mathbf{J}(\mathbf{q})\ddot{\mathbf{q}}$ and the coriolis and centrifugal term $\mathbf{f}(\ddot{\mathbf{q}}, \mathbf{q})$ can be neglected because we only consider the joint moment occurred by the weight of the support device. Lagrange multiplier vector λ can be derived as follows:

$$\begin{aligned} \lambda &= \left\{ \left[\left(\frac{\partial \mathbf{c}}{\partial \mathbf{q}}\right)^T \right]_2 \right\}^{-1} \begin{bmatrix} F_{z2'} \\ N_{2,1'} \end{bmatrix} \\ &= -\frac{1}{d + l_b} \begin{bmatrix} F_{z2'}(d + l_b)\sin\theta'_1 + N_{2,1'}\cos\theta'_1 \\ F_{z2'}(d + l_b)\cos\theta'_1 - N_{2,1'}\sin\theta'_1 \end{bmatrix} \end{aligned} \tag{12}$$

where $[(\partial\mathbf{c}/\partial\mathbf{q})^T]_2$ is an 2×2 matrix consisting of the last 2 rows of the matrix $(\partial\mathbf{c}/\partial\mathbf{q})^T$. With Lagrange multiplier vector λ and generalized moment $\tau^o = [N'_{1,2} \ N'_{2,3}]^T$, the actual joint moment of closed chain mechanism $\tau^c = [\tau^c_1 \ \tau^c_2]^T$ can be derived as follows:

$$\begin{aligned} \begin{bmatrix} \tau^c_1 \\ \tau^c_2 \end{bmatrix} &= \begin{bmatrix} N'_{1,2} \\ N'_{2,3} \end{bmatrix} - \left[\left(\frac{\partial \mathbf{c}}{\partial \mathbf{q}}\right)^T \right]^2 \lambda \\ &= \begin{bmatrix} N'_{1,2} \\ N'_{2,3} \end{bmatrix} - \begin{bmatrix} 0 & 0 \\ l'_3\cos\theta_2 & -l'_3\sin\theta_2 \end{bmatrix} \lambda \end{aligned} \tag{13}$$

where $[(\partial\mathbf{c}/\partial\mathbf{q})^T]^2$ is an 2×2 matrix consisting of the first 2 rows of the matrix $(\partial\mathbf{c}/\partial\mathbf{q})^T$.

Finally, knee joint moment caused by the weight of the device is derived as follows:

$$\tau_{\mathrm{g}} = N'_{2,3} + \frac{l'_3 F_{z2'}(d + l_{\mathrm{b}})\sin(\theta'_1 - \theta_2) + l'_3 N'_{2,1}\cos(\theta'_1 - \theta_2)}{d + l_{\mathrm{b}}} \tag{14}$$

C. Support Knee Joint Moment

To prevent the decrease in the remaining physical ability of the elderly, the support joint moment τ_{sk} is calculated as a part of the derived joint moment τ_{k}. The Support joint moment is designed based on the gravity term τ_{gra} and the GRF term τ_{GRF} from equation (3) as follows:

$$\tau_{\mathrm{sk}} = \alpha_{\mathrm{gra}}\tau_{\mathrm{gra}} + \alpha_{\mathrm{GRF}}\tau_{\mathrm{GRF}} + \tau_{\mathrm{g}} \tag{15}$$

where α_{gra} and α_{GRF} are support ratios of the gravity and GRF terms, respectively. By adjusting these ratios in the range of $0 \leq \alpha < 1$, support joint moment τ_{sk} can be determined. The gravity term τ_{gra} and the GRF term τ_{GRF} can be expressed as following equations.

$$\begin{aligned}\tau_{\text{gra}} =& m_1(\mathbf{r}_{1,\text{c}_1} - \mathbf{r}_{1,\text{c}_2} + \mathbf{r}_{2,\text{c}_2}) \times \mathbf{g} \\ &+ m_2\mathbf{r}_{2,\text{c}_2} \times \mathbf{g}\end{aligned} \tag{16}$$

$$\begin{aligned}\tau_{\text{GRF}} =& (\mathbf{r}_{1,\text{c}_1} - \mathbf{r}_{1,\text{c}_2} + \mathbf{r}_{2,\text{c}_2}) \times \sum \mathbf{f}_{\text{GRF}} \\ &- \sum (\mathbf{r}_{\text{GRF}} \times \mathbf{f}_{GRF})\end{aligned} \tag{17}$$

In the conventional control algorithm, we assumed that the term of the weight of the device τ_{g} could be neglected since we only considered support for the stance phase on flat ground. In this paper, however, we derived the knee joint moment caused by the weight of the device and add the term τ_{g} to the equation of support joint moment as shown in equation (15). By applying this algorithm to Wearable Walking Helper, it will be possible to support the user to walk considering the weight of the device. For determining the appropriate support ratios α_{gra} and α_{GRF}, we have to consider the conditions of the user such as the remaining physical ability and the disability. This is our future works in cooperating with medical doctors.

IV. Swing Phase Support Using Accelerometer

To accomplish the support of swing phase of the gait, it is necessary to measure the inclination of the link with respect to the vertical direction for calculating the support knee joint moment explained in equation (15). In this section, we first introduce a method to measure the inclination of the link with an accelerometer. Then we verify the effectiveness of the method by preliminary experiments. Finally, we apply the method to the Wearable Walking Helper with an accelerometer and evaluate the reduction of the burden of the user during gait.

A. Measuring Method of Link Inclination

As shown in Fig.4, the gravitational acceleration $g[m/s^2]$ is imposed in the vertical direction. By using the 3-axis

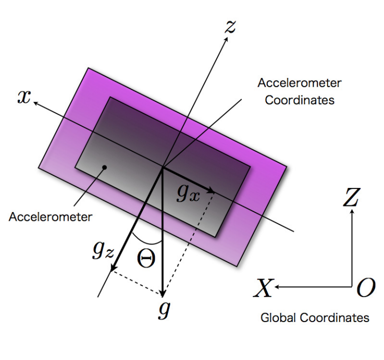


Fig. 4. Measurement of Acceleration of Human Link

accelerometer, we can measure gravitational acceleration decomposed in three directions under the condition of no dynamic acceleration. To measure the inclination of the link, we set the $x-z$ plane of the accelerometer coordinate system corresponds to the $X-Z$ plane of the global coordinate system as shown in Fig.4. Consequently, the inclination of the accelerometer can be calculated by equation (18).

$$\Theta = \tan^{-1}\left(\frac{g_x}{g_z}\right) \tag{18}$$

where Θ is inclination of the accelerometer with respect to the vertical direction. g_x and g_z are gravitational accelerations in the direction of x axis and z axis in the accelerometer coordinate system, respectively. By attaching the accelerometer to the support device, the inclination of the human links can be measured.

B. Investigation of Influence of Dynamic Acceleration

With the method for measuring the inclination proposed in the previous sub-section, we can measure the inclination of the link if dynamic acceleration does not arise. Therefore, it is necessary to investigate the influence of the acceleration arising from human motions on the accelerometer. In this section, we measure the translational acceleration of each link and investigate which links is better to attach the accelerometer for measuring the inclination of the link with respect to the vertical direction.

The measurement is conducted two motions of human: one is standing up and sitting down motions and the other is walking. To calculate translational acceleration of the links, we capture the motion of the subject by using the Motion Capture System called VICON460. Experimental results of two motions are shown in Fig.5 and Fig.6.

As shown in Fig.5, although the translational acceleration of Upper Body Link is largest, it is not so high compared to the gravitational acceleration $9.8m/s^2$. Similarly, in the case of walking experiment, the translational acceleration of Upper Body Link does not affect the measurement of the accelerometer. In the cases of the other links, the effect of the translational acceleration is too large and it must be difficult to measure the inclination of the like accurately. Especially, dynamic acceleration is highest at Foot Link during the gait, it seems impossible to measure inclination of Foot Link directly.

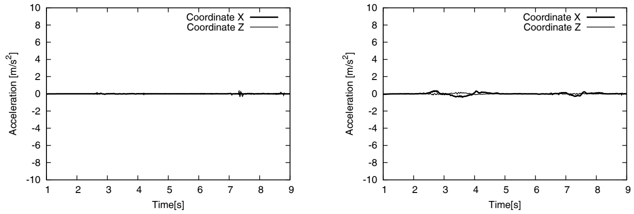
Based on these evaluations, we measure inclination of Upper Body Link instead of Foot Link. Then inclination of Foot Link θ_1 is calculated with the equation (19).

$$\theta_1 = \Theta - \theta_2 - \theta_3 - \theta_4 \tag{19}$$

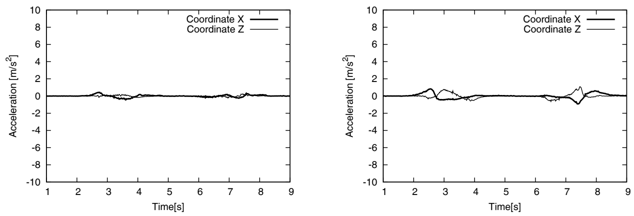
where θ_2, θ_3 and θ_4 are joint angles of ankle, knee and hip joint respectively. Θ is inclination of Upper Body Link measured with the accelerometer.

C. Preliminary Experiments

To investigate the validity of the proposed method for measuring the inclination of the link, we conducted two preliminary experiments. One is standing up and sitting down

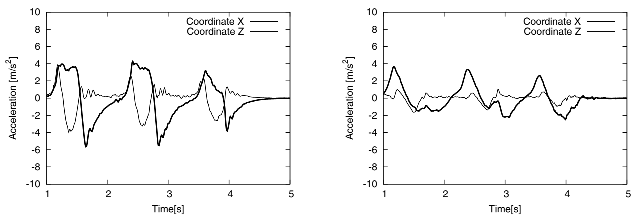


(a) Foot Link (b) Shank Link

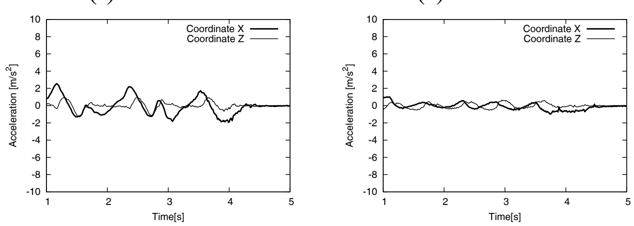


(a) Thigh Link (b) Upper Body Link

Fig. 5. Translational Acceleration During Sit-Stand Motion



(a) Foot Link (b) Shank Link

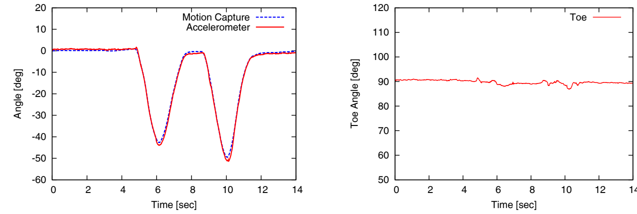


(a) Thigh Link (b) Upper Body Link

Fig. 6. Translational Acceleration During Walking

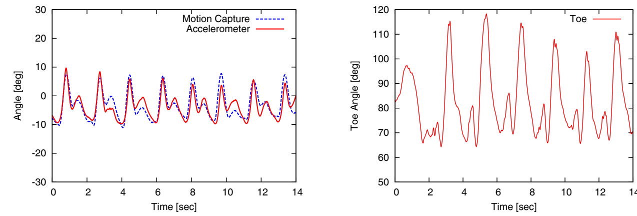
motions and the other is walking. During two experiments, we measured the inclination of the Upper Body Link using the accelerometer and joint angles using potentiometers, and then calculated the inclination of Foot Link using measured values. At the same time, we also captured the positions of markers attached to some parts of body of the user by using the Motion Capture System (VICON460) and calculated the inclination of Upper Body Link for comparing it to the measured inclination using the accelerometer.

Experimental results are shown in Fig.7 and Fig.8. As shown in Fig.7(a), inclinations with the accelerometer and Motion Capture System are almost the same. Fig.7(b) shows that inclination of Foot Link is approximately 90 degrees all through the motion. During walking, the inclination of Upper Body Link measured with the accelerometer is close to that of the value with the Motion Capture System as shown in Fig.8(a). From Fig.8(b), the inclination of Foot Link, which was conventionally assumed 90 degrees, can be calculated in real time. From these experimental results, it is possible



(a) Inclination of Upper Body (b) Inclination of Foot Link

Fig. 7. Experimental Results During Sit-Stand Motion



(a) Inclination of Upper Body (b) Inclination of Foot Link

Fig. 8. Experimental Results During Walking

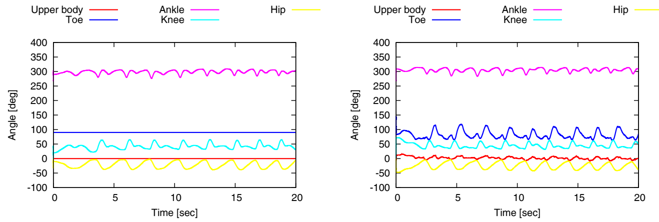
to measure inclination of Foot Link by using accelerometer, and the system could support not only the stance phase but also the swing phase of the gait appropriately.

D. Walking Experiment

The final goal of this paper is to make it possible to support not only the stance phase but also the swing phase while a user is walking. In this section, by applying the proposed method to Wearable Walking Helper, we conducted experiments to support a user during gait. To show the proposed method is effective for the reduction of burden on the knee joint, we conducted the experiments in three conditions: firstly the subject walked without support control, secondly the subject walked with only stance phase support control, and thirdly the subject walked with both stance and swing phase support control. In addition, during the experiments, we measured EMG signals of muscles conductive to the movement of the knee joint.

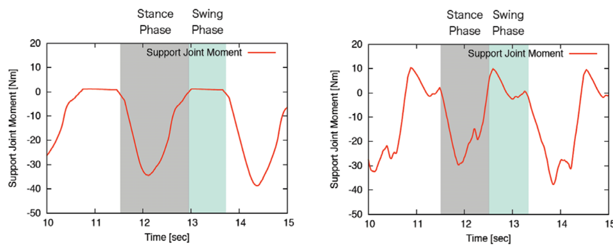
In the gait cycle, the Vastus Lateralis Muscle is active in most of the stance phase and the Rectus Femoris Muscle is active in last half of the stance phase and most of the swing phase. Therefore, during the experiments, we measure EMG signals of the Vastus Lateralis Muscle and the Rectus Femoris Muscle. The experiments were performed by the university sudedent who is 23-years-old man. Support Ratio α in the equation (15) is set to 0.6. Note that, for reducing the impact forces applied to the force sensors attached on the shoes during the gait, we utilized a low pass filter whose parameters were determined experimentally.

Joint angles during the walking experiment with only stance phase support and with both stance and swing phase supports are shown in Fig.9. Similarly, Fig.10 shows support moment for the knee joint. From Fig.9(a), the inclination of Upper Body Link was not measured and the inclination of Foot Link was unknown as the results. On the other hand,



(a) Conventional Method (b) Proposed Method

Fig. 9. Joint Angles During Walking



(a) Conventional Method (b) Proposed Method

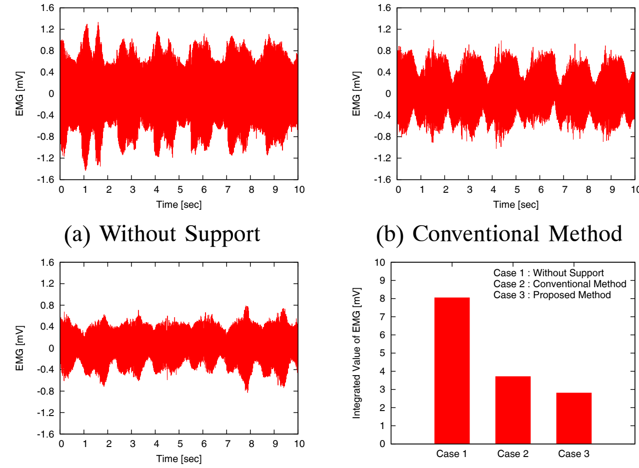
Fig. 10. Support Knee Joint Moment During Walking

with support for both stance and swing phases (Fig.9(b)), the inclination of Upper Body Link was measured by using accelerometer, and then the inclination of Foot Link was changing during the gait. From Fig.10(a), the support moment for the knee was nearly zero in swing phase with conventional method. On the other hand, with the proposed method, support moment for the knee joint was calculated and supported in both stance and swing phases as shown in Fig.10(b).

Fig.11 and Fig.12 show EMG signals of the Vastus Lateralis Muscle and the Rectus Femoris Muscle during the experiments in three conditions explained above. Fig.11(d) and Fig.12(d) are integrated values of the EMG signals. From these results, EMG signals of both the Vastus Lateralis Muscle and the Rectus Femoris Muscle have maximum values in the experiment without support and have minimum values in the experiment with both stance and swing phase supports. These experimental results show that it is possible to support both stance and swing phases by using the proposed method.

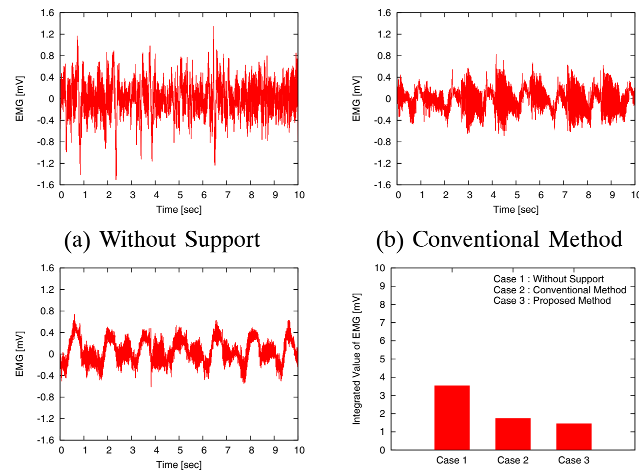
V. Conclusions

In this paper, we proposed a control method of the wearable walking support system for supporting the swing phase of the gait. In this method, we derived support moment for guaranteeing the weight of the support device and measured an inclination of the upper body of the user with respect to the vertical direction by using the accelerometer, and apply them to the method for calculating the support moment of the knee joint. The validity of the proposed method was illustrated experimentally. Further investigation and experiments based on various motions of subjects including the elderly are important on the next stage of our research. In addition, we will develop a device for supporting the both legs including the knee and hip joints.



(a) Without Support (b) Conventional Method
(c) Proposed Method (d) Integrated Values

Fig. 11. EMG Signals of Vastus Lateralis Muscle



(a) Without Support (b) Conventional Method
(c) Proposed Method (d) Integrated Values

Fig. 12. EMG Signals of Rectus Femoris Muscle

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