13

Motion Control of Wearable Walking Support System with Accelerometer Based on Human Model

Yasuhisa Hirata, Takuya Iwano, Masaya Tajika and Kazuhiro Kosuge
Department of Bioengineering and Robotics, Tohoku University
Japan

1. Introduction

Many countries of the world including Japan will become a full-fledged aged society. According to report in Japan, the elderly population aged 65 years or over in Japan will number 33 million and will account for more than 25 percent of the population. We have to support the elderly for independence in old age so that a variety of lifestyles is possible. With the development of the robot technologies, robotics researchers have developed various kinds of human assist robot such as walking aid system and manipulation aid system for supporting the elderly.

Especially, the ability to walk is one of the most important and fundamental functions for humans, and enables them to realize high-quality lives. Many researchers focused on a walker-type support system, which works on the basis of the physical interaction between the system with wheels and the user. Walkers are widely used by the handicapped because they are simple and easy to use.

Fujie et al. (1998) developed a power-assisted walker for physical support during walking. Hirata et al. (2003) developed a motion control algorithm for an intelligent walker with an omni-directional mobile base, in which the system is moved based on the user's intentional force/moment. Wandosell et al. (2002) proposed a non-holonomic navigation system for a walking-aid robot named Care-O-bot II. Sabatini et al. (2002) developed a motorized rollator. Yu et al. (2003) proposed the PAMM system to provide mobility assistance and user health status monitoring.

Wasson et al. (2003) and Rentschler et al. (2003) proposed passive intelligent walkers, in which a servo motor is attached to the steering wheel and the steering angle is controlled depending on environmental information. Hirata et al. (2007) developed The RT Walker which has passive dynamics with respect to the force/moment applied. It differs from other passive walkers in that it controls servo brakes appropriately without using any servo motors.

Many researchers have considered improving their functionality by adding wheels with actuators and controlling them based on robot technology (RT), such as motion control technology, sensing technology, vision technology, and computational intelligence. But, the size of the walker-type system is large and the user has to use the both hands for moving it. On the other hand, recently, many robotics researchers focused on wearable walking

Source: Advances in Human-Robot Interaction, Book edited by: Vladimir A. Kulyukin,
ISBN 978-953-307-020-9, pp. 342, December 2009, INTECH, Croatia, downloaded from SCIYO.COM

support system which could support the motion of the user based on the control of the actuators attached to the body of the human directly.
In U.S., performance augmenting exoskeletons has come from a program sponsored by Defense Advanced Research Projects Agency (DARPA). The goal of the program is to increase the capabilities of ground soldiers beyond that of a human (Garcia et al. (2002)). Kazeroon et al. (2006) developed Berkeley Lower Extremity Exoskeleton (BLEEX). The Sarcos Research Corporation has worked toward a full-body Wearable Energetically Autonomous Robot (WEAR) (Guizzo & Goldstein (2005)). Walsh et al. (2006) also proposed a quasi-passive exoskeleton concept which seeks to exploit the passive dynamics of human walking in order to create lighter exoskeleton devices.
Kiguchi et al. (2003), Naruse et al. (2003), Nakai et al. (2001) and Kawamoto & Sankai (2005) also developed several wearable assist systems for supporting the daily activities of the people such as walking, handling and so on. Some of the 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 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, the motions of the user could not identify correctly based on the activities of only few muscles observed by EMG signals or hardness of skin surface.
To overcome these problems, our group developed a wearable walking support system which was able to support walking activity without using biological signals (Nakamura et al. (2005)). The system calculates the support moment of the joints of the user by using an approximated human model of four-link open chain mechanism on the sagittal plane and it assists a part of the joint moment by the actuator of the wearable walking support system. In the conventional control algorithm, however, we assumed that the system only supported the stance phase of the gait and we neglected the weight of the support device in the stance phase. We also assumed that the user stood on flat ground and inclination of Foot Link is always parallel to the ground.
When we consider the support of the swing phase of the gait, 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 chapter, we derive the support moment for the knee joint to guarantee the weight of the device. We also propose a method for measuring the inclination of a link of the human model with respect to the vertical direction by using an accelerometer. By using these methods, we derive the support moment of the joint for supporting the user in not only the stance phase but also the swing phase. We applied the proposed methods to the developed wearable walking support system experimentally and the experimental results illustrate the validity of them.

2. Wearable Walking Helper

In this section, we briefly introduce a developed wearable walking support system called Wearable Walking Helper. We developed the smaller and lighter support device for the knee joint than its conventional system proposed by Nakamura et al. (2005). 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.

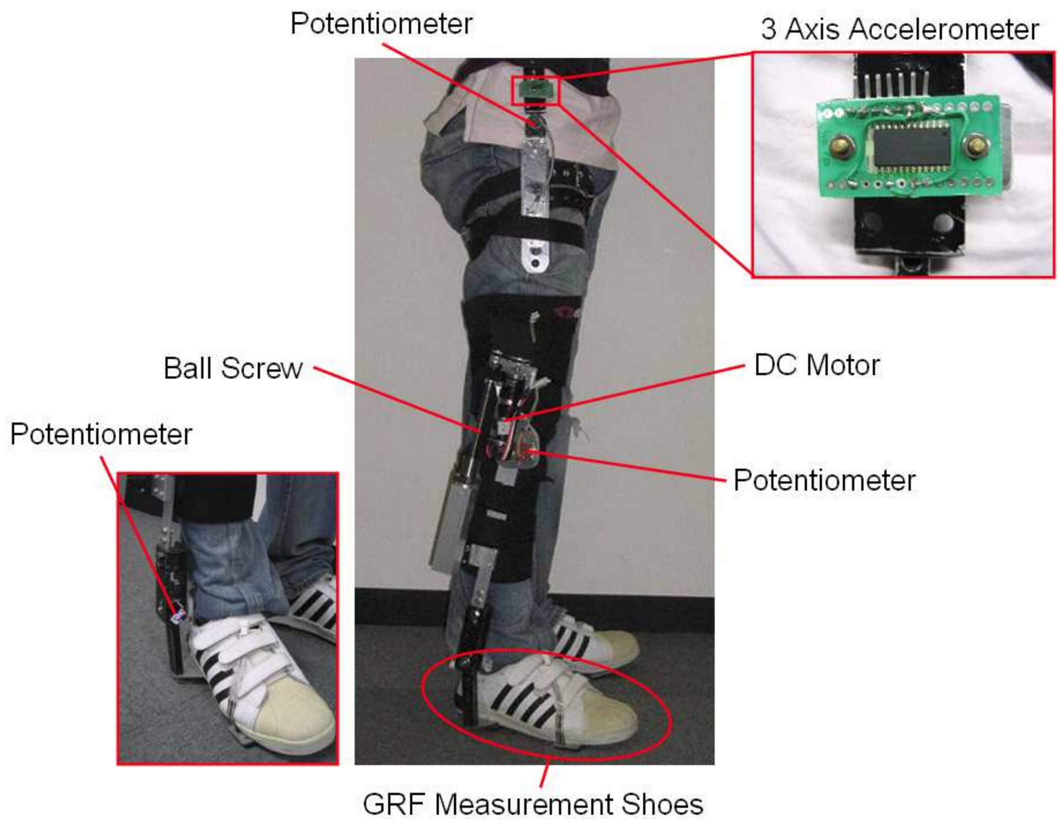


Fig. 1. Wearable Walking Helper with Accelerometer

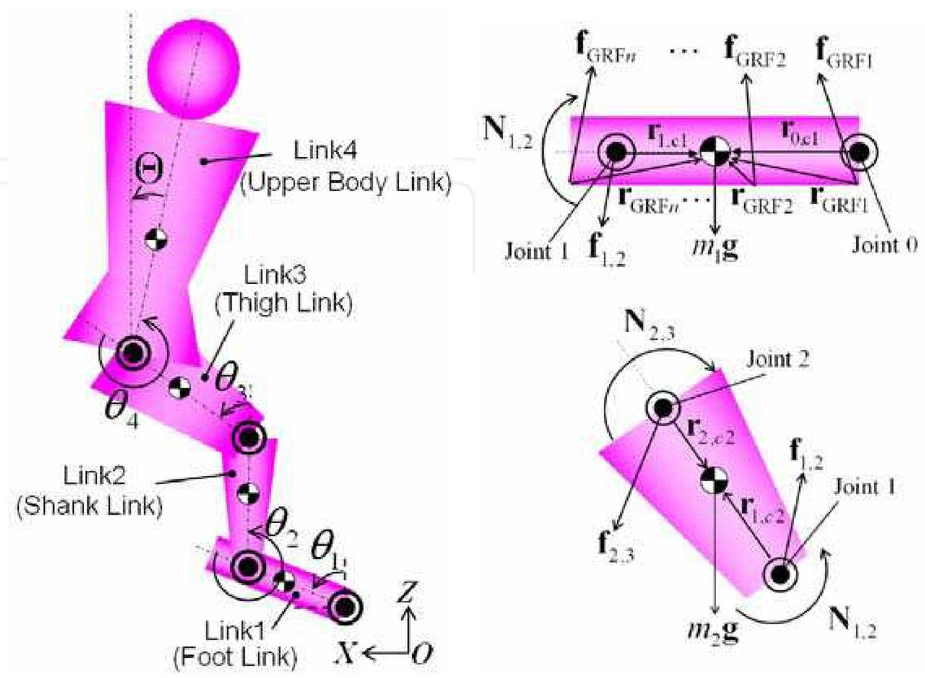
The system has three potentiometers attached to the ankle, knee and hip joints to measure the rotation angle of each joint. To measure the Ground Reaction Force (GRF), we utilize two force sensors attached to the shoe sole: one is on the toe and the other is on the heel. In addition, a 3-axis accelerometer is attached to near the hip joint to measure inclination of the link. By using measured joint angles, GRFs and the inclination of the link, the control algorithm proposed in this chapter calculates the support moment around the knee joint.

3. 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 also drive the knee joint moment caused by the weight of the device itself. Finally, we determine the support joint moment to be generated by the actuator of the support device.

3.1 Calculation of knee joint moment using human model

To control the Wearable Walking Helper, we use the 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:

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

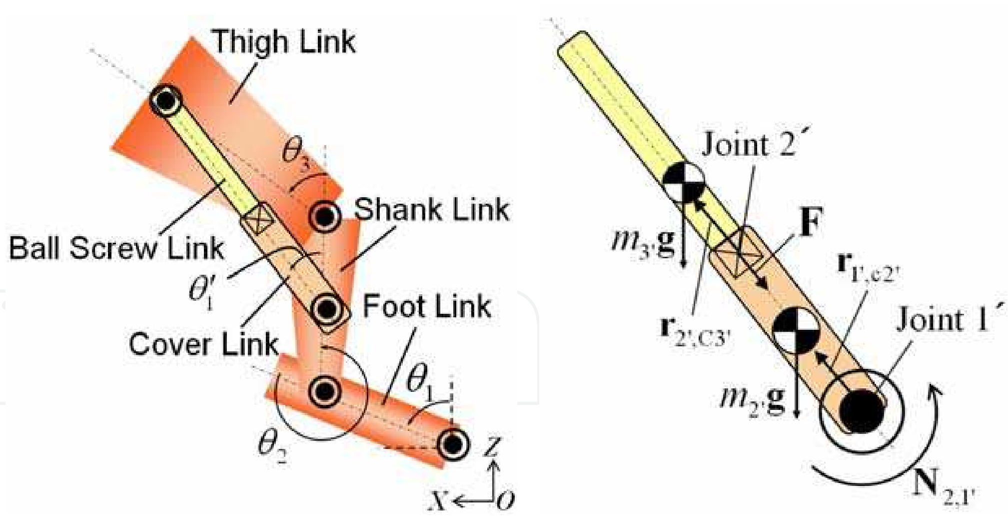
$$N_{i-1,i} - N_{i,i+1} + \mathrm{r}_{i,c_i} \times \mathrm{f}_{i,i+1} - \mathrm{r}_{i-1,c_i} \times \mathrm{f}_{i-1,i} = I_i \frac{d\dot{\theta}_i}{dt} \tag{2}$$

where, $\mathrm{f}_{i-1,i}$ and $\mathrm{f}_{i,i+1}$ are reaction forces applying to the joint i and $i+1$ respectively. m_i is the mass of the link i, g is the vector of gravity acceleration and $\dot{\mathrm{v}}_{c_i}$ 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:

$$\begin{aligned}\tau_{\mathrm{k}} = N_{2,3} = &-I_1 \frac{d\dot{\theta}_1}{dt} - I_2 \frac{d\dot{\theta}_2}{dt} - m_1(\mathrm{r}_{1,c_1} - \mathrm{r}_{1,c_2} + \mathrm{r}_{2,c_2}) \times (\dot{\mathrm{v}}_{c_1} - \mathrm{g}) - m_2 \mathrm{r}_{2,c_2} \times (\dot{\mathrm{v}}_{c_2} - \mathrm{g}) \\ &+ (\mathrm{r}_{1,c_1} - \mathrm{r}_{1,c_2} + \mathrm{r}_{2,c_2}) \times \sum \mathrm{f}_{\mathrm{GRF}} - \sum (\mathrm{r}_{\mathrm{GRF}} \times \mathrm{f}_{\mathrm{GRF}})\end{aligned} \tag{3}$$

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



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

Fig. 3. Device Model

3.2 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, we could not derive the joint moment easily. A variety of schemes for deriving joint torques for robots consisting of closed chain mechanisms have been proposed by Luh & Zheng (1985), Nakamura (1989) and so on. In this research, we apply the method proposed by Luh & Zheng (1985).

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:

$$\begin{aligned} N_{2,3}^{'} = & -I_1^{'}\frac{d\dot{\theta}_1}{dt} - I_2^{'}\frac{d\dot{\theta}_2}{dt} - I_{2'}\frac{d\dot{\theta}_{2'}}{dt} - m_1^{'}(\mathrm{r}_{1,c_1}^{'} - \mathrm{r}_{1,c_2}^{'} + \mathrm{r}_{2,c_2}^{'}) \times (\dot{\mathrm{v}}_{c_1}^{'} - \mathrm{g}) \\ & - m_2^{'}\mathrm{r}_{2,c_2}^{'} \times (\dot{\mathrm{v}}_{c_2}^{'} - \mathrm{g}) - m_{2'}\mathrm{r}_{2',c_2} \times (\dot{\mathrm{v}}_{c_{2'}} - \mathrm{g}) - m_{3'}\mathrm{r}_{3',c_2} \times (\dot{\mathrm{v}}_{c_{3'}} - \mathrm{g}) \end{aligned} \tag{4}$$

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

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

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

where $\mathrm{F}=[F_{x2'}F_{z2'}]^T$ 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

$$A_{1'}^{2}\,A_{2}^{3'}=\begin{bmatrix}\cos\theta_2 & 0 & \sin\theta_2 & l_3'\sin\theta_2\\ 0 & 1 & 0 & 0\\ -\sin\theta_2 & 0 & \cos\theta_2 & l_2'+l_3'\cos\theta_2\\ 0 & 0 & 0 & 1\end{bmatrix}=\begin{bmatrix}\mathrm{R}_{1'2}^{23'} & \mathrm{P}_{1'2}^{23'}\\ 0 & 1\end{bmatrix} \tag{7}$$

and similarly from the joint 1′ to the joint 3′ through the joint 2′ is

$$A_{1'}^{2'}\,A_{2'}^{3'}=\begin{bmatrix}\cos\theta_1' & 0 & \sin\theta_1' & d\sin\theta_1'+l_b\sin\theta_1'\\ 0 & 1 & 0 & 0\\ -\sin\theta_1' & 0 & \cos\theta_1' & d\cos\theta_1'+l_b\cos\theta_1'\\ 0 & 0 & 0 & 1\end{bmatrix}=\begin{bmatrix}\mathrm{R}_{1'2'}^{2'3'} & \mathrm{P}_{1'2'}^{2'3'}\\ 0 & 1\end{bmatrix} \tag{8}$$

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.

$$\mathrm{c}=\mathrm{P}_{1'2}^{23'}-\mathrm{P}_{1'2'}^{2'3'}=\begin{bmatrix}l_3'\sin\theta_2-(d+l_b)\sin\theta_1\\ l_2'+l_3'\cos\theta_2-(d+l_b)\cos\theta_1'\end{bmatrix}=\begin{bmatrix}0\\0\end{bmatrix} \tag{9}$$

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

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

where

$$\left(\frac{\partial\mathrm{c}}{\partial\mathrm{q}}\right)^T=\begin{bmatrix}0 & 0\\ l_3'\cos\theta_2 & -l_3'\sin\theta_2\\ -\sin\theta_1' & -\cos\theta_1'\\ -(d+l_b)\cos\theta_1' & (d+l_b)\sin\theta_1'\end{bmatrix} \tag{11}$$

Additionally, in the equation (10), the inertia term $\mathrm{J(q)\ddot{q}}$ and the coriolis and centrifugal term $\mathrm{f(\ddot{q},q)}$ 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:

$$\lambda=\left\{\left[\left(\frac{\partial\mathrm{c}}{\partial\mathrm{q}}\right)^T\right]_2\right\}^{-1}\begin{bmatrix}F_{z2'}\\ N_{2,1'}\end{bmatrix}=-\frac{1}{d+l_b}\begin{bmatrix}F_{z2'}(d+l_b)\sin\theta_1'+N_{2,1'}\cos\theta_1'\\ F_{z2'}(d+l_b)\cos\theta_1'+N_{2,1'}\sin\theta_1'\end{bmatrix} \tag{12}$$

where $[(\partial c/\partial q)^T]^2$ is an 2 × 2 matrix consisting of the last 2 rows of the matrix $(\partial c/\partial q)^T$. With Lagrange multiplier vector λ and generalized moment $\phi^o = [N'_{1,2}\ N'_{2,3}]^T$, the actual joint moment of closed chain mechanism $\phi^c = \left[\tau_1^c\ \tau_2^c\right]^T$ can be derived as follows:

$$\begin{bmatrix}\tau_1^c \\ \tau_2^c\end{bmatrix} = \begin{bmatrix}N'_{1,2} \\ N'_{2,3}\end{bmatrix} - \left[\left(\frac{\partial c}{\partial q}\right)^T\right]^2 \lambda = \begin{bmatrix}N'_{1,2} \\ N'_{2,3}\end{bmatrix} - \begin{bmatrix}0 & 0 \\ l'_3\cos\theta_2 & -l'_3\sin\theta_2\end{bmatrix}\lambda \tag{13}$$

where $[(\partial c/\partial q)^T]^2$ is an 2× 2 matrix consisting of the first 2 rows of the matrix $(\partial c/\partial q)^T$.
Finally, knee joint moment caused by the weight of the device is derived as follows:

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

3.3 Support knee joint moment

To prevent the decrease in the remaining physical ability of the elderly, we calculate the support joint moment τ_{sk} as a part of the derived joint moment τ_k. The joint moment expressed by equation (3) consists of the gravity term τ_{gra} and the GRF term τ_{GRF}. Therfore, we calculate the support joint moment as follows:

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

where α_{gra} and α_{GRF} are support ratios of the gravity and GRF terms, respectively. By adjusting these ratios in the range of $0 \le \alpha < 1$, 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.

$$\tau_{gra} = m_1(r_{1,c_1} - r_{1,c_2} + r_{2,c_2}) \times g + m_2 r_{2,c_2} \times g \tag{16}$$

$$\tau_{GRF} = (r_{1,c_1} - r_{1,c_2} + r_{2,c_2}) \times \sum f_{GRF} - \sum (r_{GRF} \times f_{GRF}) \tag{17}$$

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 the Wearable Walking Helper, it could support 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 disabilities. This is our future works in cooperating with medical doctors.

4. Swing phase support using accelerometer

To accomplish the support of swing phase of the gait, the system has to detect 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.

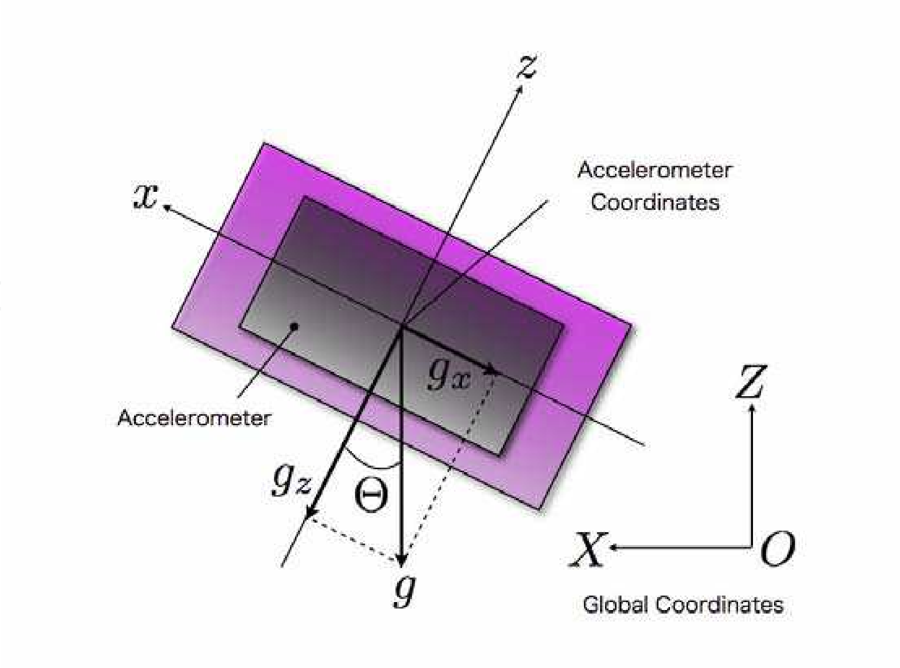


Fig. 4. Measurement of Acceleration of Human Link

4.1 Measuring method of link inclination

As shown in Fig. 4, the gravitational acceleration $g[m/s^2]$ is imposed along the vertical direction. By using the 3-axis accelerometer, the system can measure the 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 system can calculate the inclination of the accelerometer by using the following equation:

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

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 system can measure the inclination of the human links.

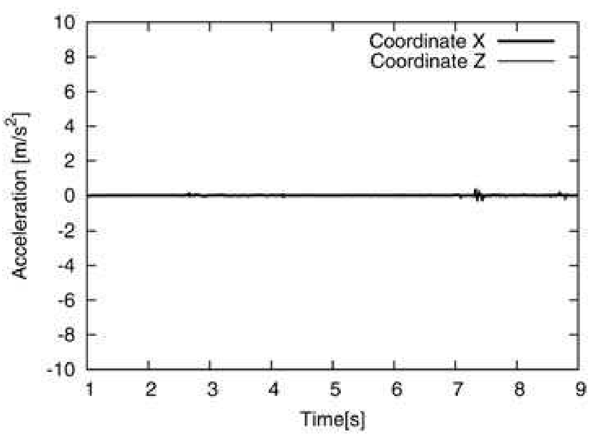
4.2 Investigation of influence of dynamic acceleration

With the method for measuring the inclination proposed in the previous section, we can measure the inclination of the link if dynamic acceleration does not arise. Therefore, we should 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.
In the measurement experiment, we conducted two motions of human: one is standing up and sitting down motions and the other is walking. To calculate translational acceleration of

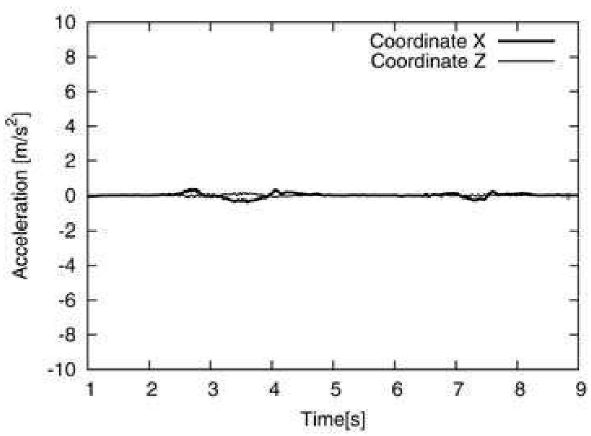
the links, we captured the motion of the subject by using the Motion Capturing System called VICON460. Fig. 5 and Fig. 6 shows the experimental results of two motions.
As shown in Fig. 5, although the translational acceleration of Upper Body Link is largest, it is not so high compared to the gravitational acceleration 9.8 $[m/s^2]$. 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 the inclination of Foot Link directly.
Based on these evaluations, we decided to measure the inclination of Upper Body Link instead of Foot Link. Then inclination of Foot Link θ_1 is calculated with the following equation:

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

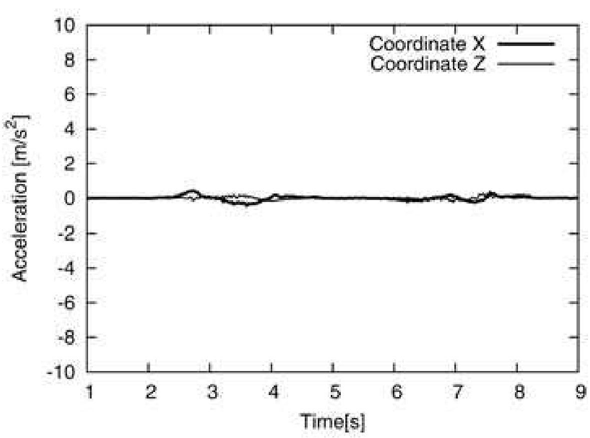
where θ_2, θ_3 and θ_4 are joint angles of ankle, knee and hip joint respectively. Θ is inclination of Upper Body Link measured with the accelerometer.



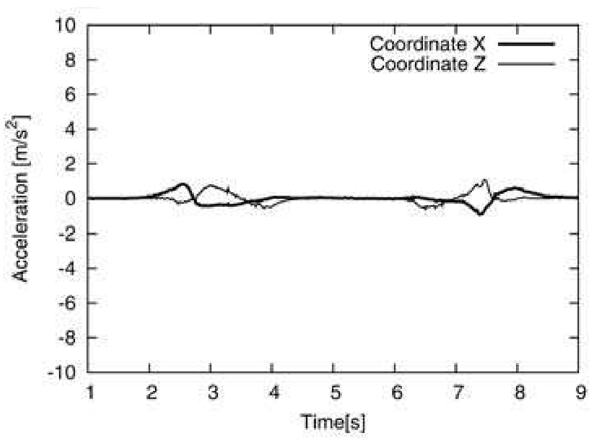
(a) Foot Link



(b) Shank Link

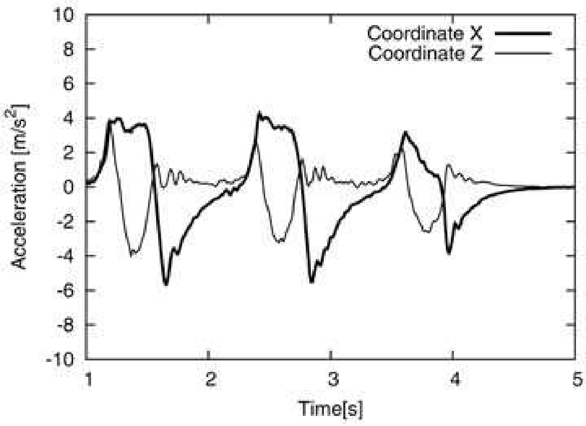


(a) Thigh Link

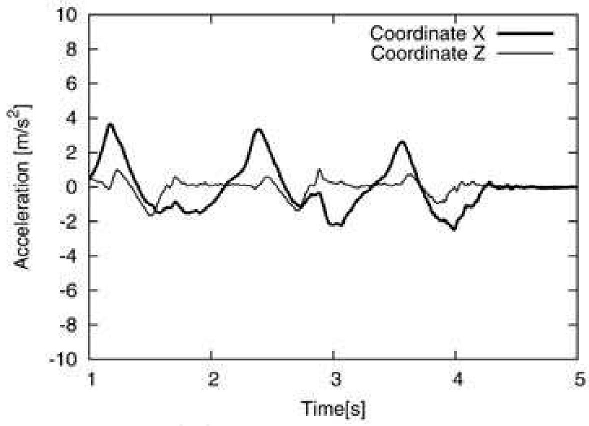


(b) Upper Body Link

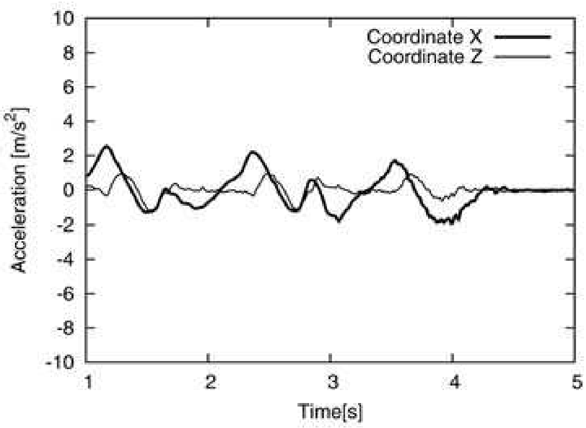
Fig. 5. Translational Acceleration During Sit-Stand Motion



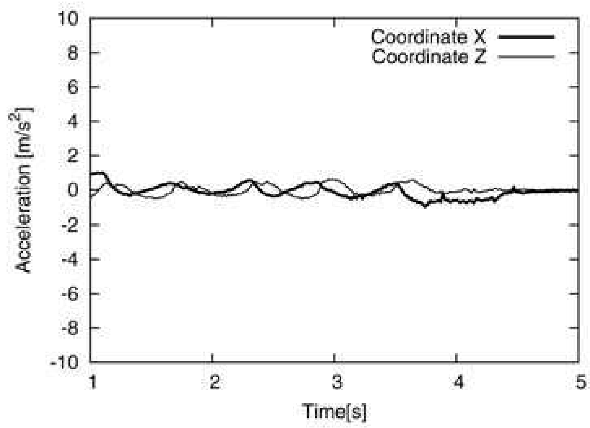
(a) Foot Link



(b) Shank Link



(a) Thigh Link



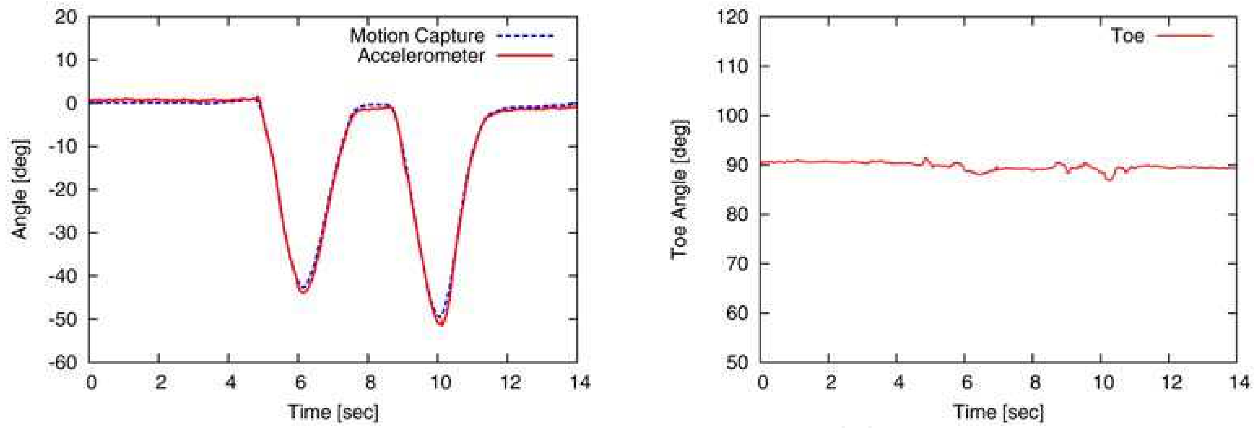
(b) Upper Body Link

Fig. 6. Translational Acceleration During Walking

4.3 Preliminary experiments

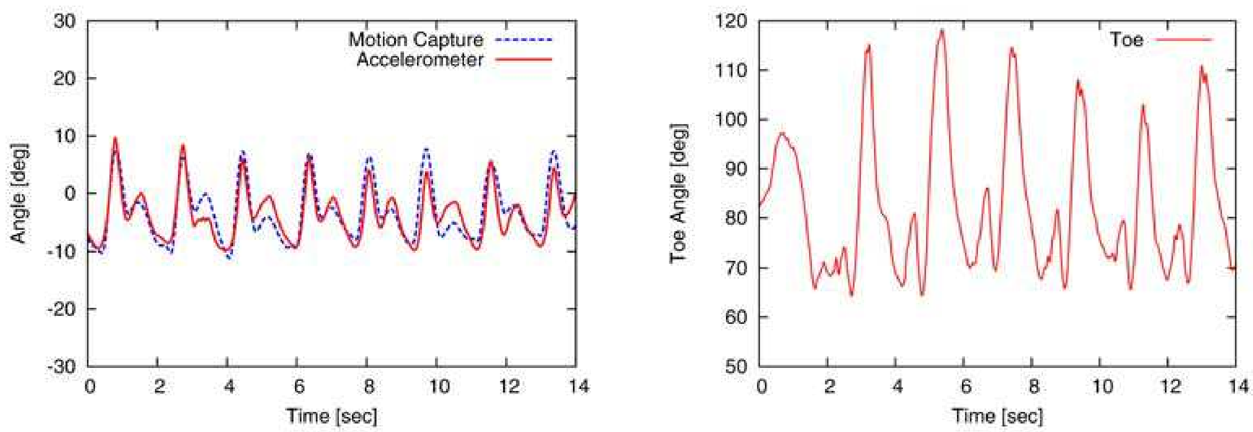
To investigate the validity of the proposed method for measuring the inclination of the link, we conducted two preliminary experiments. One is standing up and sitting down motions and the other is walking. During two experiments, we measured the inclination of the Upper Body Link using the accelerometer 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 Capturing 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 Capturing 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 Capturing 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, you can see that we 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.



(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

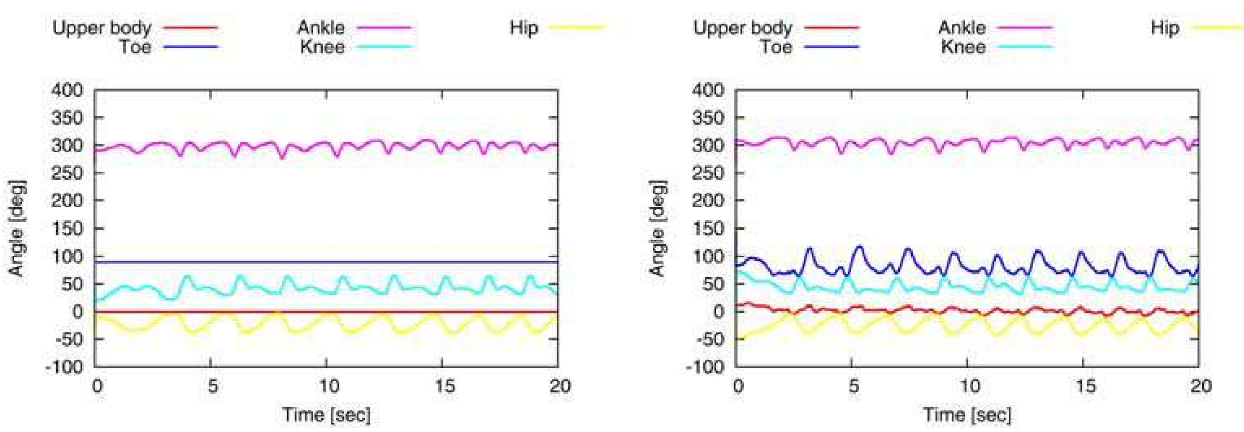
Fig. 8. Experimental Results During Walking

5. 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.

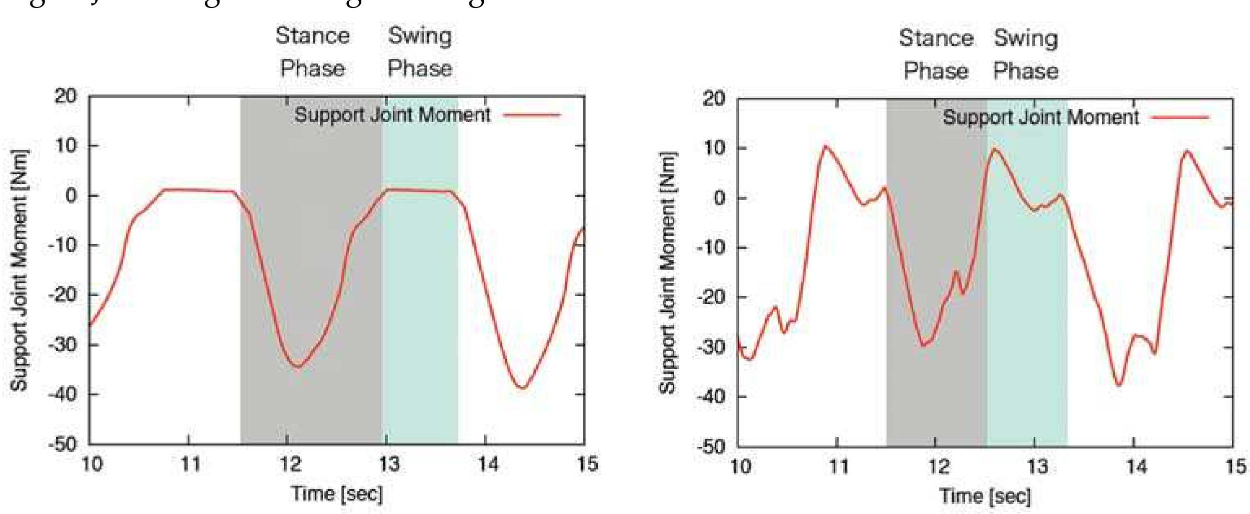
In the gait cycle, the Vastus Lateralis Muscle is active in most of the stance phase and the Rectus Femoris Muscle is active in last half of the stance phase and most of the swing phase. Therefore, during the experiments, we measured EMG signals of the Vastus Lateralis Muscle and the Rectus Femoris Muscle. The university student who is 23-years-old man performed the experiments. Support Ratio α_{gra} and α_{GRF} in the equation (15) were set to 0.6, respectively. 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, 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 system changed the inclination of Foot Link 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).



(a) Conventional Method (b) Proposed Method

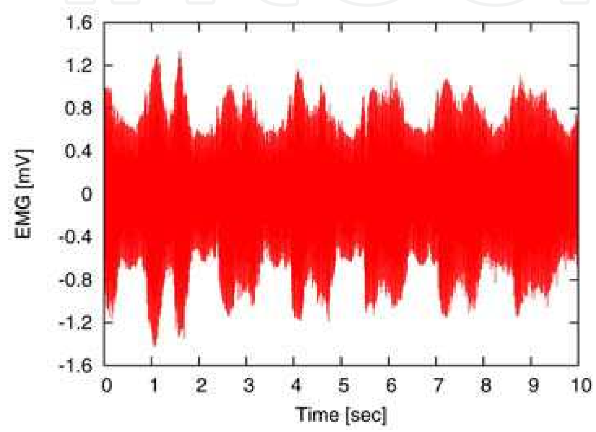
Fig. 9. Joint Angles During Walking



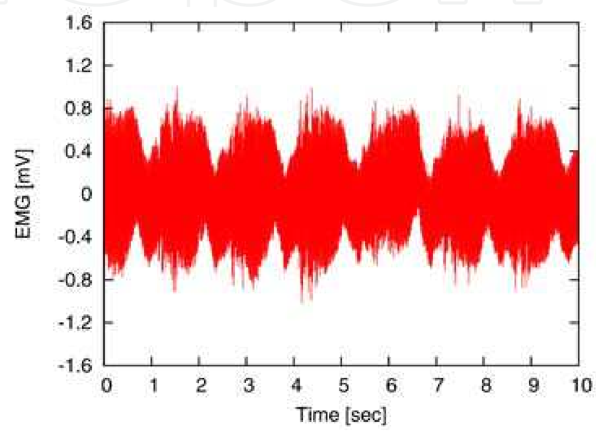
(a) Conventional Method (b) Proposed Method

Fig. 10. Support Knee Joint Moment During Walking

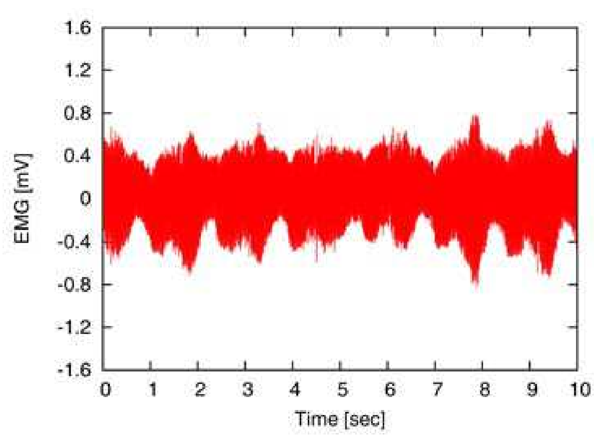
Fig. 11 and Fig. 12 show EMG signals of the Vastus Lateralis Muscle and the Rectus Femoris Muscle during the experiments in three conditions explained above. Fig. 11(d) and Fig. 12(d) shows the integrated values of the EMG signals. From these results, EMG signals of both the Vastus Lateralis Muscle and 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 the developed system can support both stance and swing phases.



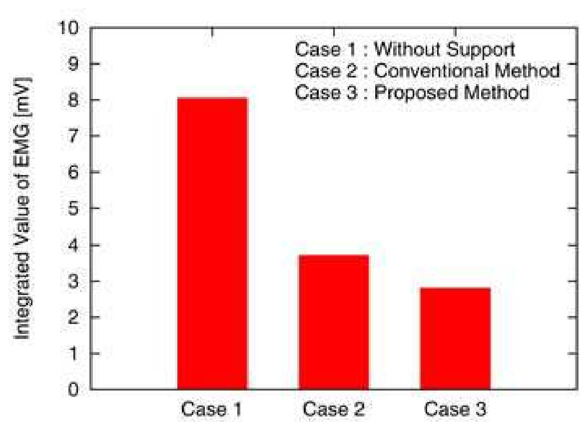
(a) Without Support



(b) Conventional Method

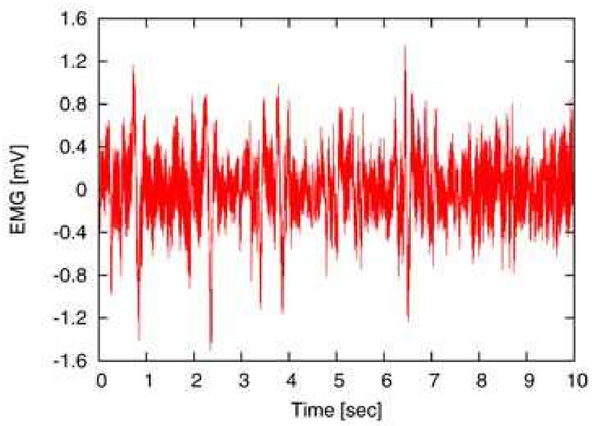


(c) Proposed Method

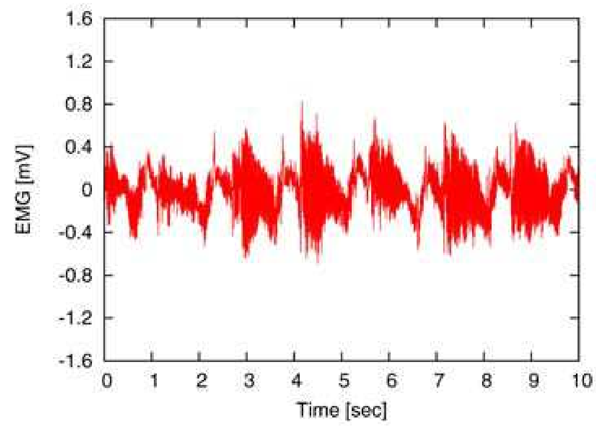


(d) Integrated Values

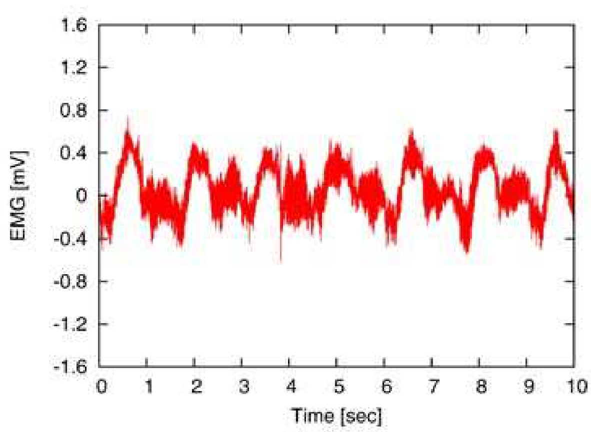
Fig. 11. EMG Signals of Vastus Lateralis Muscle



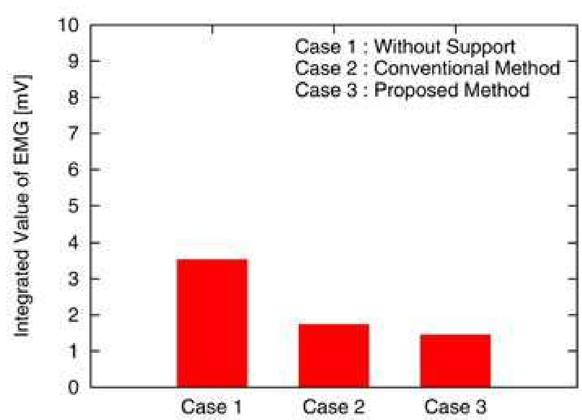
(a) Without Support



(b) Conventional Method



(c) Proposed Method



(d) Integrated Values

Fig. 12. EMG Signals of Rectus Femoris Muscle

6. Conclusions

In this paper, we proposed a control method of the wearable walking support system for supporting not only the stance phase but also 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. We applied 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.

7. References

Fujie, M., Nemoto, Y., Egawa, S., Sakai, A., Hattori, S., Koseki, A., Ishii, T. (1998). Power AssistedWalking Support andWalk Rehabilitation, In: *Proc. of 1st InternationalWorkshop on Humanoid and Human Friendly Robotics*

Hirata, Y., Baba, T., Kosuge, K. (2003). Motion Control of Omni-directional typeWalking Support System "Walking Helper", In: *Proc. of IEEE Workshop on Robot and Human Interactive Communication*, 2A5

Wandosell, J.M.H., Graf, B. (2002). Non-Holonomic Navigation System of a Walking-Aid Robot, In: *Proc. of IEEE Workshop on Robot and Human Interactive Communication*, 518-523

Sabatini, A. M., Genovese, V., Pacchierotti, E. (2002). A Mobility Aid for the Support to Walking and Object Transportation of People with Motor Impairments, In: *Proc. of IEEE/RSJ Intl. Conf. on Intelligent Robots and Systems*

Yu, H. Spenko, M., Dubowsky, S. (2003). An Adaptive Shared Control System for an Intelligent Mobility Aid for the Elderly, In: *Auton. Robots*, Vol.15, No.1, 53-66

Wasson, G., Sheth, P., Alwan, M., Granata, K., Ledoux, A., Huang, C. (2003). User Intent in a Shared Control Framework for Pedestrian Mobility Aids, In: *Proc. of the 2003 IEEE/RSJ Intl. Conf. on Intelligent Robots and Systems*

Rentschler, A. J., Cooper, R. A., Blaschm, B., Boninger, M. L. (2003). Intelligent walkers for the elderly : Performance and safety testing of VA-PAMAID robotic walker, In: *Journal of Rehabilitation Research and Development*, Vol. 40, No. 5

Hirata, Y., Hara, A., Kosuge, K. (2007). Motion Control of Passive Intelligent Walker Using Servo Brakes, In: *IEEE Transactions on Robotics*, Vol. 23, No.5, 981-990

Garcia, E., Sater, J. M., Main, J. (2002). Exoskeletons for human performance augmentation (EHPA): A program summary, In: *Journal of Robotics Society of Japan*, Vol. 20, No. 8, 44-48

H. Kazerooni et al. (2006). The Berkeley Lower Extremity Exoskeletons, In: *ASME J. of Dynamics Sys., Measurements and Control*, V128

Guizzo, E., Goldstein, H. (2005). The rise of the body bots, In: *IEEE Spectrum*, Vol. 42, No. 10, 50-56

Walsh, C. J., Pasch, K., Herr, H. (2006). An autonomous underactuated exoskeleton for loadcarrying augmentation, In: *Proc. IEEE/RSJ International Conference on Ingellignet Robots and Systems*, 1410-1415

Kiguchi, K. Tanaka, T., Watanabe, K., Fukuda, T. (2003). Exoskeleton for Human Upper-Limb Motion Support, In: *Proc. of IEEE ICRA*, 2206-2211

Naruse, K. Kawai, S. Yokoi, H. Kakazu, Y. (2003). Development of Wearable Exoskeleton Power Assist System for Lower Back Support, In: *Proc. of IEEE/RSJ IROS*, 3630-3635

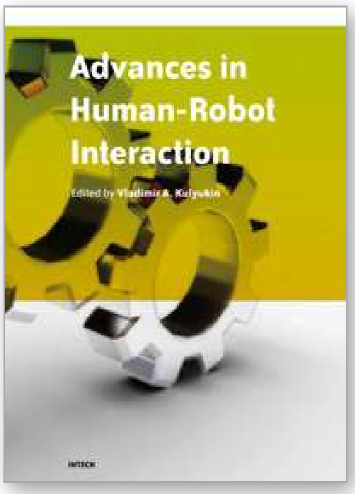
Nakai, T. Lee, S, Kawamoto, H., Sankai, Y. (2001). Development of Power Assistive Leg for Walking Aid using EMG and Linux, In: *Proc. of ASIAR*, 295-299

Kawamoto, H., Sankai, Y. (2005). Power Assist Method Based on Phase Sequence and Muscle Force Condition for HAL, In: *Advanced Robotics*, Vol.19, No.7, 717-734

Nakamura, T, Saito, K., Kosuge, K. (2005). Control ofWearableWalking Support System Based on Human-Model and GRF, In: *Proc. of IEEE ICRA*, 4405-4410

Luh, J., Zheng, Y.F. (1985). Computation of Input Generalized Forces for Robots with Closed Kinematic Chain Mechanisms, In: *IEEE J. of Robotics and Automation*, 95-103

Nakamura, Y. Ghodoussi, M.?(1989). Dynamics computation of closed-link robot mechanisms withnonredundant and redundant actuators, In: *IEEE Transactions on Robotics and Automation*, Vol.5, No.3, 294-302



Advances in Human-Robot Interaction
Edited by Vladimir A. Kulyukin

ISBN 978-953-307-020-9
Hard cover, 342 pages
Publisher InTech
Published online 01, December, 2009
Published in print edition December, 2009

Rapid advances in the field of robotics have made it possible to use robots not just in industrial automation but also in entertainment, rehabilitation, and home service. Since robots will likely affect many aspects of human existence, fundamental questions of human-robot interaction must be formulated and, if at all possible, resolved. Some of these questions are addressed in this collection of papers by leading HRI researchers.