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A Gait Support System for Human Locomotion without Restriction of the Lower Extremities: Preliminary Mechanism and Control Design

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Abstract— **A motion support system is important for improving the quality of life for people with inadequate muscle power for walking or standing. Such a system must maintain coordination between the movement of the living body and that of the motion support system with respect to timing and power. We have developed a preliminary assistance device for human locomotion that does not restrict movement of the user's lower extremities. To determine the appropriate timing and applied force for this assistance device, we have use two types control mode; (1) surface EMG signals from the lower limb in preliminary controller. Each assistance leg has a three-axis magnetic sensor near lower end which tracks a magnet attached to each ankle of the user's legs. This sensor attracted to a magnet attached to the human ankle in the leg swing phase. The mechanical leg supports the user's weight when the leg is in the standing phase.**

I. Introduction

A motion support system is an important way to improve the quality of life for people who have inadequate muscle power for walking or standing. Kazerooni et al. developed a human power amplifier (BLEEX) which used positive feedback with inverse kinematics of the exoskeleton system, and supports the movement of the extremities [1, 2]. Sankai et al. developed a power assistance device for the lower limbs (HAL-3) which used an electromyogram (EMG) information [3]. Liu et al. designed lower extremity leg exoskeleton controlled by foot reaction force and measurement angles using wearable mechanical link with encoders [4]. Although these devices were successfully used to control the assistance limbs in sync with human legs, they could not eliminate contact force between the human limb and the mechanical limb. This limitation resulted in the obstruction of limb movement and locomotion because of the connection between the assistance system and the extremity. An important aspect of the development of motion support systems was coordinating the movement of the human body and that of the motion support system with respect to timing and power. Furthermore, the characteristics of impedance and the viscosity with regard to the movement of the joint were very important factors in determining the optimum power and timing of the device. However, accurately estimating the characteristics of neuronal signal processing, muscle, and joints of the human body was difficult. Using inexact parameter values to control an assistance limb with a rigid connection to a human limb could be lead to obstructed movement of the limb or the uncomfortable sensation that the user's own limb is being driven by the assistance device.

In view of such difficulties, we were developing a gait support system which was not connected to the legs. This system was intended for use by people who had experienced a reduction in muscle power in order to assist them with walking, standing, and lifting. This support system was not connected to the person through the legs, so the user did not have any obstruction or inaccurate coordination between the movement of the leg and that of mechanical system. Such concept was reported in virtual reality research [5, 6]. Yoshikawa et al. reported an effectiveness of an encounter type haptic display which contacted a human finger when the human finger was positioned virtual surface. These systems could decrease discomfort experience in their movement.

The design of the proposed gait support system and its controller used two type modes to determine the power and timing of the assistance force; a) ground reaction force model, b) surface EMGs from the leg muscles. This article reports

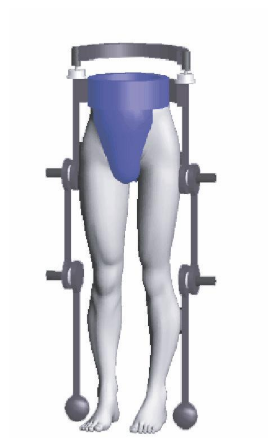

Fig. 1. A schematic drawing of the gait support system for human locomotion. There are two degrees of movement, at the hip and knee, on each assistance leg. The support system and the operator have a rigid connection at the pelvis only, not through the legs.

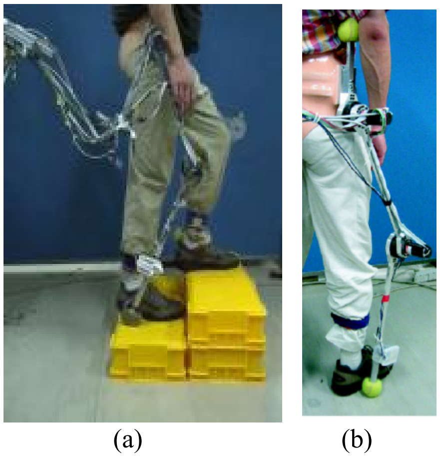

Fig. 2. . (a) The operator wore a gait support system while climbing stairs. (b) back view of the system. This system has two degrees of movement in each leg and is driven by a 150 W DC motor. A three-axis magnetic sensor (white box) is attached to the end of the assistance leg. A magnet was attached to each ankle (blue band). The assistance legs were attached to the pelvis.

the efficiency of the preliminary mechanical design and control system in experiments.

II. Gait Support System for Human Locomotion

A. Mechanical design

Our gait support system was designed for elderly people with redu ced muscle strength who are able to walk but need a support device such as a cane in order to walk or stand for an extended period. We focused on the function of a cane as a weight-bearing device which depends on the user's upper body. Then we designed the gait support system that functions as automatic walking sticks. The schematic design is shown in Figure 1. This system has two degrees of movement in each leg, at the hip joint and the knee joint. Each joint has a 150 W DC motor with a 1000 ppr encoder and a Harmonic Drive gear. A three-axis magnetic sensor (white box in Figure 2) is attached to the end of each mechanical leg. Another end of the leg was rigidly attached to the hip attachment (blue part). The attachment was connected to the ischial tuberosity and pelvis. This device is not connected to the legs. The assistance legs work as walking sticks attached to the sides of the operator's body.

B. Control of the support

The control of the support system were contained 4ch motor drivers with a 12 bit D/A, 8ch EMG amplifier with a 12bit A/D, 8ch force sensor with 12bit A/D, and 6ch magnetic sensor with 12 bit A/D, and 4ch encoder counter (Figure 3). The sampling frequency of the controller PC was 1 kHz. The 8 ch EMG signals were used to determinate the assistance force or to evaluate the efficiency of the gait support system.

There were two basic control phases in this system: the swing phase and the stance phase. In the swing phase, the mechanical assistance leg, with the three-axis magnetic

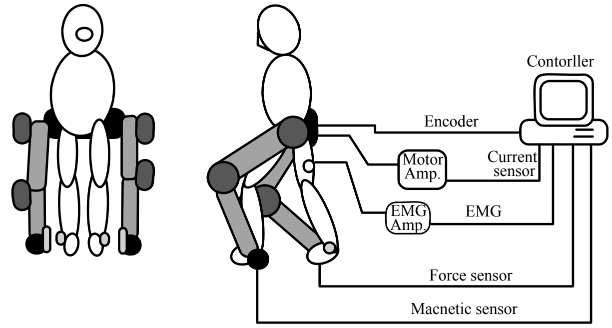

Fig. 3. Schematic drawing of the assistance legs and their control system. This system consists of four motor controllers and a measuring apparatus for EMG, force, and magnetic sensor signals.

sensor, traces the path of the operator's ipsilateral leg via a Neodymium magnet attached to the operator's ankle. Therefore, the mechanical leg was not obstructed in the swing phase, and the trajectory and landing point of the leg were determined by the operator without conscious control. The three-dimensional position of the magnet on the ankle was calculated using the following magnetic dipole model:

$$\begin{aligned}\mathbf{H} &= -gradU_m \\ &= -\nabla\left(\frac{\mathbf{M}\bullet\mathbf{r}}{4\pi\mu_0 r^3}\right) \\ &= \frac{1}{4\pi\mu_0 r^3}\left(3M\frac{\mathbf{r}}{r}\cos\theta - \mathbf{M}\right)\end{aligned} \quad (1)$$

where U_m and $\boldsymbol{H}$ are the magnetic potential and magnetic field at the sensor position, respectively (point O in Figure 4); $\boldsymbol{M}$ is the magnetic moment of a source magnet, the strength of which is M; $\boldsymbol{r}$ is the position vector between the source magnet and the magnetic sensor; and μ_0 is the magnetic permeability.

In the stance phase, the mechanical legs bore the weight of the payload and the operator's body to reduce the impact force and gravitational force on the skeleton during locomotion. The assistance force was determined by the kinematical method or by estimation of the surface EMG signals on the lower legs, as explained below.

C. Determination of assistance force and timing

We used two type methods to determine the assistance force and timing.

In Method 1, the ground reaction force (GRF) that the stance leg bore was calculated based on kinematics using positional information of human limb. The force gradually increased to maximum in the early period of stance phase. It decreased when the stance leg had transitioned to the swing phase. This calculated GRF was applied in the assistance controller to determine the assistance force.

In Method 2, EMG signals from the lower limb were used to determine the assistance force. The joint torque of the body was correlated with the EMG activity of the connecting muscles [7]. In order to reduce the impact on the skeleton and joints, we then determined the onset timing of the assistance force before the leg met the ground. We focused on the EMG activity and pre-activation before the stance phase to detect the onset timing. The muscles of the human body have a shock-absorbing function that is activated before the foot touches down. Visual feedback helps to determine touchdown timing [8]. Therefore, this pre-activation occurs as a result of information obtained from the physical senses to predict contact with an external surface. We tried to make predictions based on EMG signals. The medial vastus muscle was activated at 100 ms before touchdown to absorb the impact flection force. Principal component analysis (PCA) was employed to detect this activation using EMG signals from both sides of the medial vastus muscle and the anterior tibial muscle:

where Z_j is the principal component and v_{ij} and x_i are the coefficient and input data vector, respectively. $|v_{ij}|$ was used for the index which represents each muscle activity. When muscle activity increases, the corresponding $|v_{ij}|$ increases because of the increasing variance of the corresponding input.

$$Z_j = \sum_i v_{ij} x_i \qquad (2)$$

III. Experiments

A. Follow-up control of the swing leg

The trajectories of the human leg and the mechanical leg were measured while a subject was walking on a treadmill with a gait support system. The subject walked at 2.5 km/h. The position of the magnetic sensor was calculated by inverse kinematics of the mechanical leg, and the position of the source magnet was measured using the magnetic dipole model as described above.

B. Gait support

Effectiveness analysis was performed by using a force plate which measured the floor reaction force. When the subject (65 kg), using the gait support system, walked on the force plate, the reaction force under the human leg was measured with (Method 1) and without support.

C. EMG analysis

Characteristics of pre-activation signals on EMG during usual walking were investigated to determine the amplitude

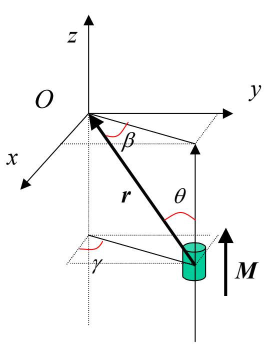

Fig. 4. Magnetic dipole model to determine the three-dimensional position of source magnet M. The three-axis magnetic sensor is point O.

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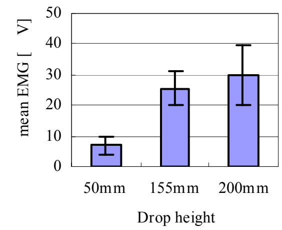
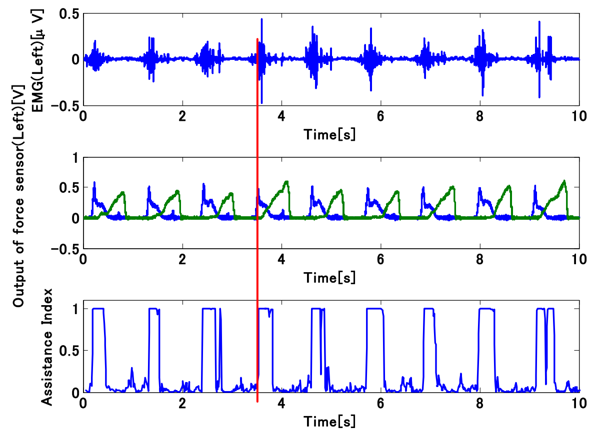


Fig. 7. Mean amplitude of pre-activation of the medial vastus muscle before touchdown with eyes open. Pre-activity depended on falling height.



A. Trajecto

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deviation betw... about (x, y) = (30, 20) mm in the swing phase. This result shows that the assistance leg followed the human leg using the magnet dipole model and the operator controlled the standing point of the assistance leg unconsciously.

B. Gait support

Figure 6 shows the effectiveness of the gait support system during walking. With gait support, the reaction force was reduced by about 140 N in the impact phase and by about 90

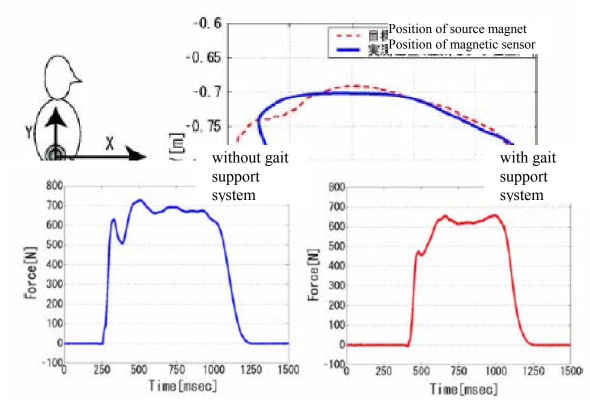

Fig. 6. Reactive force of human leg during walking (Method 1).

N in the midstance phase. The operator was assisted with appropriate force, without being hindered by the mechanism, because there was little difference in the shape of the time cource of the reactive force between with and without support and the amplitude of the reactive force was reduced.

The controller could not respond smoothly to unsteady locomotion, because the controller is based on steady-state walking on a flat surface. This control system does not have enough information to recognize human intent. In the real world, a person naturally adjusts his or her steps and reaction force to avoid falling down when walking on unevev surfaces or climbing stairs.

C. EMG analysis

As shown in Figure 7, the pre-activation of the medial vastus muscle before touchdown when the subject's eyes were open was correlated with the falling height. This suggests that visually obtained information about a person's surroundings is coded in the pre-activation by some controller of human body. This information might be used to reduce the impact force when a human walks on an even surface or climb stairs.

Figure 8 shows the assistance index calculated as explained above, and compares this index with touchdown timing measured by force sensor on the sole of the subject's foot and from the EMG of the medial vastus muscle while the subject walked on treadmill without the support system. The index increased almost as soon as the swing leg touched down. This index might therefore be applicable for controlling the onset timing of the transition from the swing to the stance phase. Estimation of the touchdown onset and earlier contact control using the information before human leg touchdown are necessary to reduce the impact force on the human skeleton and joints. This onset information should contribute to reducing the impact force when our gait support system is used. The efficiency of this determination method of assistance force is investigating in execution.

V. CONCLUSION

The preliminary design and basic function of a gait support system that does not hold the lower extremity have been demonstrated. The system assists human locomotion without disturbance and reduces the reaction force on the operator. Then we proposed the method to determine when and where the operator's leg would be touchdown using surface EMG from lower extremity. This proposed control system might be applied much more situation at locomotion. Next step of this study, the gait support system with this controller based on EMG signal will be applied to diversified environment, and its efficiency will be studied.

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Fig. 8. EMG of the medial vastus muscle, force sensor on the human sole, and calculated assistance index to control onset timing of assistance force. The index increased almost as soon as the swing leg touched down.