Lateral Balance Supporting Device for Postural Reflex 
Ambulatory Experiments

Tytus WOJTARA¹, Makoto SASAKI¹, Shingo SHIMODA¹, Fady ALNAJJAR¹, and Hidenori KIMURA¹

¹RIKEN

Abstract— The human balance, the ability to keep ones body upright while standing or walking, deteriorates in old age or can be compromised after accidents or brain surgeries. With the aged society, age related balance problems are on the rise. Persons with balance problems are more likely to fall during their everyday life routines. Especially in elderly, falls can lead to bone fractures making the patient bedridden, weakening the body and making it more prone to other diseases. Health care expenses for a fall patient are often very high. There is a great deal of research being done on exoskeletons and power assists. However, these technologies concentrate mainly on the amplifications of human muscle power while balance has to be provided by the human themself.

Our research has been focused on supporting human balance in harmony with the human's own posture control mechanisms such as postural reflexes. This paper proposes an artificial balancer that supports human balance through acceleration of a flywheel attached to the body. Appropriate correcting torques are generated through our device based on the measurements of body deflections. In this paper we present ambulatory experiments on a balance beam. These experiments have demonstrated the effectiveness of our device in supporting balance while walking and the possibility of enhancing balance-keeping capability in human beings through the application of external torque.

I. INTRODUCTION

Recently, postural reflex attracts interest from many researchers due to the coming aged society, [1] [2]. In biomechanics, balance is an ability to maintain the center of gravity of a body within the base of support with minimal postural sway as defined in [3]. It is the ability to resist the disrupting gravitational force. Elderly people gradually lose this ability due to various factors like sensory deterioration, concentration capacity decrease, weakening brain performance and muscles etc. It is reported that the three sensory modalities relevant for balance, that is, proprioception, vestibular organs and vision decline in old age, [4], [5]. Also, the number of Purkinje cells in the cerebellum diminishes which causes increased tremor and body sway preventing smooth movements [6]. The reaction time is longer and the muscles become weaker. Since walking and balancing require a lot of concentration, it is difficult to carry out other tasks simultaneously as reported in [7]. In case of accident victims or people after brain surgeries, parts of the brain or the nervous system are damaged and balance-keeping ability is impaired. An excellent review on the balance recovery process of stroke patients is found in [8].

Patients with balance problems are prone to falls even while carrying out their everyday life routines. In case of hip joint fractures the patients become bedridden which often leads to other diseases. Medical costs for bedridden patients are very high. Preventing falls in everyday life would be a great contribution to society. Persons with balance impairments use tools like canes, crutches, walkers, joint stiffening orthoses etc. These tools usually reduce the number of degrees of freedom. A great deal of research is done on exoskeletons and power assists like those described in [9], [10] and [11]. However, all of these projects concentrate only on the muscle power amplification, not on the balance itself. Since biped walking is a combination of locomotion and balance, balance assistance is also important in exoskeletons. This paper proposes a balance support device for elderly or persons with postural reflex impairments.

In order to support balance, we use a flywheel attached to the body to externally apply counter-torque against the disrupting gravitational torque to the human body.

There are inverted gyro-pendulum, bicycle or mono-cycle balance controls using flywheels, for example [12]. However, applying the same principle to the human body is much more challenging. The human body is made of soft tissue and a rigid attachment of a flywheel is difficult.

When designing a controller, the human balance control system, whether functional or impaired, must be taken into account.

In contrast to our research, gyro stabilizers that use mass rotating at constant speed are tested as human balance support.

Research on Functional Electrical Stimulation (FES) or Functional Neuromuscular Stimulation (FNS) is progressing which is the electrical stimulation of muscles to provoke their movement [13]. This can be invasive, requiring an operation, or non-invasive but only targeting muscles close to skin surface. This method is mostly used to activate paralyzed muscles that otherwise wouldn’t move. For our purpose, however, it seems neither practical nor reliable.

The device we propose in this paper is purely non-invasive. It uses the rotational action/reaction principle of the flywheel attached to the human body through a tight and stiff corset.
We have developed a feedback control algorithm for applying rotational torque to counterbalance the body against disrupting torque, that is consistent with and compliments to human innate postural reflex.

We carried out ambulatory experiments to prove the usefulness of our device and algorithms. In ambulatory experiments, the subject was walking on a balance beam. During ambulation, our device was supporting the lateral balance. The results show that our device reduced the range of body sway and other parameters during ambulatory experiments.

II. POSTURE REFLEX AND SUPPORT DEVICE

A. Balance-keeping as Feedback Control

A person about to lose his lateral balance, usually uses his arms to regain equilibrium as shown in Fig. 1a. Also, the movement of the upper body relative to the lower body plays a big role, but here we want to concentrate on the arm movement only. The person rotates the arms around the shoulder joints in the sagittal plane by applying torque. This action generates a reaction torque of the same size but opposite direction acting on the upper body. The torque is applied as long as the arms accelerate. If, in this short time, the torque is sufficient to regain the body balance, the arms can be decelerated. The torque produced during this deceleration is absorbed by the body. If the torque was not sufficient to regain the balance in time, the angle range of the arms rotation ends and the arms have to stop, which means decelerate. This produces torque which forces the body even more out of balance. In such a case the person can rescue its balance only by stepping or jumping aside. A 170 cm tall 70 kg heavy standard human’s both arms (each 75 cm long, 3.5 kg heavy and the center of mass at the distance of 40 cm from the axis of rotation, the shoulder joint) have a moment of inertia $1.12 \text{kgm}^2$ around the corresponding shoulder joints as calculated from the biometric data in [14].

The balance-keeping action described above is considered to be a feedback mechanism that generates a counter torque based on the detection of the body deflections and/or the disturbance torques. Although the detailed neural mechanism of balance-keeping is still not known, it can be represented in the block diagram of Fig. 2. It has already been experimentally demonstrated that the three sensory modalities, proprioceptive, vestibular and visual, play a big role in ambulation and detecting disrupting torques on the body. It is the most interesting question how these three independent sensory modalities are integrated by the CNS (Central Nervous System) to generate correcting motions. Also, it is our daily observation that the correcting motions are generated by the torso. It is not known how an appropriate combination of muscle movements is chosen and integrated against specific disturbances, although some hypothesis are proposed [15], [16] and [17].

B. Artificial Balancer

Tight wire artists (funambulists) exhibit an astonishing capability of balance-keeping. However, instead of relying on their arms that have relatively low moment of inertia, they use a long pole with a high moment of inertia as shown in Fig. 1b. Since torque $\tau$ is the product of moment of inertia $I$ and the angular acceleration $\alpha$, i.e. $\tau = I \alpha$, a funambulist can produce high torque, even with small accelerations of the pole. The disadvantage of the pole is the same as the of human arms, and namely, a limited angle range ($\pm 90\text{deg}$). Also, since it usually is very long it requires a large space for maneuvers. Limits of an angle range become a problem if the balance cannot be regained before running into them.
Our device uses the same principle as human arms or tight wire pole, however the correcting torque is produced by the flywheel, instead of arms or pole, attached to the back as shown in Fig. 1c. The torque is applied to the upper body and generated by the reaction of the angular acceleration of the flywheel. Compared with the pole, it has an advantage of unlimited angle range and compact size.

The flywheel rotation is determined by a feedback control law based on the deflection of the body that is detected by a gyro sensor attached to the chest. The feedback control law was designed so as to smoothly help the human innate balance-keeping mechanism.

To avoid misunderstandings, it has to be noted that the gyro effect is not used, since for our application, it is a parasitic effect that has to be avoided. Therefore, when the flywheel’s support is not needed, it should be gradually decelerated so as not to apply any disturbing torques.

The flywheel has no angle range limits, but it has an angular speed limit (as the human arms and the tight wire pole have). For security reasons and because of the parasitic gyro effect, we limited the angular speed of the flywheel to 3000rpm. When the speed gets limited, no torque can be produced any more in the same direction. The time integral of the torque \( \tau(t) \) is limited by a given moment of inertia \( I \) and angle speed limit \( v_{\text{max}} \) as expressed in Eq. 1, where \( t \) denotes the time.

\[
\int \tau(t) \, dt \leq I v_{\text{max}}
\]  

The torque itself, or the angular acceleration, is limited by the parameters of the motor. For control, the reference roll angle was set before commencing the experiment by measuring the offset between the vertical and the angle perceived as vertical by the subject. This offset angle was then subtracted from the measured angle value.

Our device consists functionally of three parts: a sensor, an actuator and a controller. The actuator is an electrically driven flywheel. The sensor is a posture sensor. Both the actuator and the sensor are attached to the human body by a corset. The controller and the power supply are not worn by the human at this stage of hardware development.

The human wears a hip corset which is made of resin foam. A flywheel made of aluminum is attached to the corset. Most of the flywheel’s mass is located far from its center of rotation to maximize the moment of inertia. The flywheel can freely rotate around the human antero-posterior axis. An electric motor can accelerate or decelerate the flywheel in both directions. The motor is a flat type brush-less DC motor with hall sensors for commutation. The motor is coupled with the flywheel by a timing belt in a way that reduces speed and increases torque. Some parameters of the flywheel system are listed up in Table 1.

The flywheel is connected to the corset at hip hight. This hight revealed to be the most comfortable while the weight is least felt by the human. The gyro sensor 3DM-GX1 is a product of MicroStrain and has an on-board microcomputer to process and send data via RS232.

The controller is a servo motor controller equipped with a resistor to absorb back EMF during deceleration of the massive flywheel. The algorithm is implemented on a PC equipped with digital-to-analog converters, pulse counters and RS232 ports. The desired torque signal is sent from the PC to the servo controller as an analog signal. The data from the attitude sensor is received by RS232 circuit at the PC.

The rotating parts are protected by an acrylic glass cover to minimize danger of injury.

### C. Control Law

Postural reflexes are automatic movements that control the equilibration against disrupting environmental inputs embedded in brain motor control. As an add-on external support, our artificial balancer must be consistent with these human postural reflexes.

We use an attitude sensor to measure the roll angle of the upper body. This is the angle around the antero-posterior axis of the human upper body \( \theta_2 \), also denoted \( \theta \) since it’s the only measured angle. The lower body angle is denoted by \( \theta_1 \) as in Fig. 3a. The angles are zero when the body stands vertically.

Even a standing human exhibits body sway or postural sway. This sway varies among individuals and is in the range \( \pm 1\,\text{deg} \) as described in [18]. Also the speed of the sway shows individual variations. Since the body sway is natural and necessary for balance-keeping, our control algorithms don’t interfere as long as the body sway size and speed are below the threshold.

Standing in shoulder deep water, it is easy to keep balance. The water helps to keep balance due to its viscosity. It acts as a damper or brake. The disadvantage is however, that after the balance is lost, it is difficult to regain it due to the same viscosity effect that helps to keep balance. We realized damping when falling, but no damping when returning to the equilibrium position.

If the upper body angle is above threshold the algorithm interprets it as being off balance. The algorithm, we propose, incorporates this and is mathematically described in Eq. 2. Figure 3b describes it in the phase plane.

<table>
<thead>
<tr>
<th>parameter name</th>
<th>value</th>
</tr>
</thead>
<tbody>
<tr>
<td>flywheel weight</td>
<td>2.2kg</td>
</tr>
<tr>
<td>flywheel diameter</td>
<td>300mm</td>
</tr>
<tr>
<td>flywheel moment of inertia</td>
<td>0.028kgm²</td>
</tr>
<tr>
<td>motor stall torque</td>
<td>4.5Nm</td>
</tr>
<tr>
<td>motor nominal torque</td>
<td>0.5N/m</td>
</tr>
<tr>
<td>motor maximal permissible speed</td>
<td>5000rpm</td>
</tr>
<tr>
<td>pulley ratio</td>
<td>4 : 1</td>
</tr>
</tbody>
</table>
\[ \tau(\theta, \dot{\theta}) = \begin{cases} k_d \left( \dot{\theta} - \frac{\dot{\theta}}{|\dot{\theta}|} \dot{\theta} D \right) & \theta \dot{\theta} > 0 \text{ and } |\theta| > \dot{\theta} D \\ 0 & \text{else} \end{cases} \] (2)

The area where the human posture control is assumed to work correctly is described as "natural recovery area". The parameters were set to: upper body angular speed threshold \( \dot{\theta} D = 0.03 \text{deg/s} \), controller D-gain \( k_d = 60 \text{Nm/deg} \). Those parameters were tuned through questioning of the subject to achieve most convenient support function.

The advantage of this controller is that it only supports human balance if the human fails to do so by himself. It therefore saves energy used for balance support.

The dead zone in which control is not active is set up in order to allow for natural body posture fluctuations and is represented as a rectangle of lengths \( 2\theta P \) and \( 2\dot{\theta} D \) with center at the origin. The areas P, D and PD are those where only, respectively, P-control, D-control or PD-control are active.

III. Balance Beam Ambulatory Experiments;

Feedback

All experiments were carried out with the permission from RIKEN Ethics Committee. The 10 subjects who took part in the experiments were in the age range 26 – 35 years, body weight range 62 – 80 kg and body height range 161 – 180 cm.

Experiments were carried out to verify our device's balance support capacity during human ambulation. A balance beam was chosen for ambulatory experiments because walking on a balance beam challenges lateral balance, while antero-posterior stability is high. The beam is 1700 mm long, 50 mm wide and 110 mm high. The subject was asked to walk the entire length without shoes in small steps putting one foot in front of the other in a distance of few centimeters, as shown in Fig. 1d. The subjects were told to walk at their comfortable pace but not to hurry if they start to lose balance and rather recover in place. Subject's arms were closed to avoid arm balancing, making the data evaluation easier. During the trials, a closed loop controller with the algorithm as in Eq. 2 was used. Two cases, support (Sup) and disturbance (Dis) were applied. In case of disturbance the same magnitude but opposite polarity torque was applied.

The results in Fig. 4 show maximal body sway measured at the neck. It shows mean values and the standard deviations of total 43 trials done by 8 subjects.

The number or stepping down was counted. It was 0.05 times for (Sup) and 1.48 for (Dis) case. The mean time needed to go over the beam was 6.83 s for (Sup) and 8.67 s for (Dis) case. The time is shown in the figure as relative time to person’s mean traverse time.

IV. Discussion

In balance beam ambulatory experiments we investigate the potential of our device to help more complex and continuous voluntary movements. The experiments demonstrated the potential of our device for helping daily life of the elderly.

The control law as in Eq. 2 is good for some subjects but not for others who might like other control laws. Tailoring specific control law to each subject must have been necessary for obtaining sharper results, if we take into account the individuality of postural reflex. Also, since we only used a single gyro sensor for detecting body sway, we were not able to measure other parts of body movements, e.g. leg deflection, which may play important roles in postural reflex. Despite lower effectiveness the t-tests showed a statistical difference between disturbance and support case.

The results of ambulatory experiments, as in Fig. 4, show that our device reduced the human body sway (left), the time to traverse the beam (middle) and the number of stepping down (right). Also according to subjects, it was easier to keep balance while walking over the balance beam with a support. Some subjects were able to cross the balance beam smoother and in shorter time if supported.

Ambulatory experiments do not have an external disturbance. The maximal upper body sway mean diminished but the difference is not statistically significant. The reason for that is probably in the fact that subjects use different strategies. While some subjects try to keep their balance by swaying extensively, others avoid large sways by stepping...
down from the beam or touching the floor with their toes. This again implies the necessity of tailoring of feedback control for each subject.

The number of stepping down, including touching the floor was almost zero during support and increased during disturbance. It should be noted that all subjects reported without any reservation that they got the better feeling of balance enhancement when they were supported by our equipment.

The success of balance recovery is not only dependent on the size of the torque applied to the body but also strongly depends on the detection or prediction of instability, which is not the main topic of this paper.

V. Conclusions

We have developed an artificial balancer, a balance-keeping device that supports postural reflexes. We experimentally showed that it is possible to counteract the body disturbance by directly applying external torque to the body. We also showed that it is possible to enhance the balance-keeping capability of humans through reinforcing postural reflex by an add-on feedback control mechanism.

Our device and its experimental verification of balance enhancement power shown in this paper open up a new research direction of postural reflex through construction of balance enhancers. Our device has a potential to create a new postural reflex for balance-keeping through exercise, which gives a new way of rehabilitation for the disabled. Also, combining our balance enhancer to the exoskeleton which is a sort of muscle enhancer, will give a new possibility to prosthesis.

Our results presented in this paper mark only the preliminary stage of the future long process of developing practical artificial balancers. Our artificial balancer deals only with lateral movements and its total weight has to be reduced so that the device can be used by physically challenged persons. Motor, battery and other parts can be included in the rotor and so contribute to the moment of inertia keeping the total mass of the device light weight. A more efficient feedback control law that seamlessly matches the human’s innate postural reflex must be designed. At this moment, however, we have only very limited information about the control aspect of postural reflex.

We are collaborating with a medical institution and aiming to produce a device that will be used in clinical practice, supporting patients with impaired balance and in rehabilitation to train their balance ability. The device can also be used together with an exoskeleton and so extend its application field.

REFERENCES


