An Effective Assistive Device for Decreasing Resistant Forces Caused by Walking

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Abstract The purpose of this study is to develop a simple back assistive device that generates sufficient assistive moment and is not subject to resistant forces while walking. It is widely known that many labor workers suffer from low back pain. To diminish their low back pain, development of a simple and practical back assistive device for use in the workplace is highly recommended. Some simple back assistive devices using elastic elements have been proposed. However, under walking conditions, elastic tension generated by flexion of the hip joint exerts a resistant force against walking. Since the majority of physical workers in industries or other fields have to walk while performing tasks, they require a back assistive device that has sufficient assistive moment in a bending forward posture and does not generate a resistant force while walking. The problem is how to decrease the undesirable resistant force from elastic elements caused by walking, while maintaining a back-supporting property in bending forward tasks. Focusing on the reciprocal motion of legs in walking, we devised a simple mechanism by which elastic tension is diminished under walking conditions. On the other hand, the tension of elastic elements is available in bending forward posture. Making use of the reciprocal motion of legs, two swing arms mounted on the device rotate synchronously with legs. Using this mechanism, elastic tension is almost completely eliminated under walking conditions. The swing arms rotate in the same direction under walking condition, but they cannot rotate in the opposite direction, so that elastic elements are stretched to produce an assistive moment for the back in a bending forward posture. The device proved to be effective in reducing undesirable tension in the lower belt under walking condition. The tension of elastic elements caused by walking decreased to less than one-quarter compared with that of conventional back assistive devices.

Keywords: back assistive device, elastic element, resistant force canceller, parallelogram.

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1. Introduction

In many occupations, workers often experience low back pain. Many caregivers leave their job due to low back pain. Biomechanical evaluations of various assistive devices for transferring nursing home residents from a bed to a chair have been reported [1]. Such devices reduce biomechanical stress, thereby decreasing the occurrence of low back injuries related to handling of the residents. However, there is still considerable room for improvement of the assistive devices. Abdoli-E et al. [2] developed a simple on-body back assistive device (PLAD) using elastic elements. The PLAD significantly reduced muscular effort of lumbar and thoracic erector spinae. The amount integrated EMG reduction ranged from 14.4% to 27.6% for the lumbar and thoracic erector spinae, respectively. This reduction may prevent the risk of recurring back injuries especially in repetitive tasks. The same research group further confirmed that the PLAD decreased the 3D dynamic moments on lumbar [3] and back muscular fatigue during a repetitive lifting task [4]. They reported that the highest

PLAD tension elicited the greatest reduction in erector spinae activity [5]. It is interesting to note that wearing the PLAD does not alter the oxygen consumption during a free-style lifting and lowering task. They also found no differences between oxygen consumption during PLAD and no PLAD conditions [6]. It should be noted that all the experimental results were obtained from simple repetitive lifting tasks. In actual working situations, however, users of the devices often have to walk while performing tasks. From the biomechanical point of view, the PLAD is really effective in reducing lower back loading. However, there may be a side effect. The elastic tension of PLAD acts as a resistant force under walking condition.

A Cochrane literature review reported by Verbeek et al. [7] aimed to examine the effectiveness of training and provision of assistive devices in preventing and treating low back pain in workers exposed to manual material handling. Surprisingly, they concluded that training workers to use proper manual material handling techniques and providing them with assistive devices are not effective interventions in preventing low back pain. This fact shows the difficulty of developing back assistive devices that can effectively prevent low back pain. We have attempted for years to develop a back assistive device that elicits small resistant force while walking [8]. This paper introduces a simple idea by which a back assistive device can be used with almost no resistant force caused by walking.

2. Method

Figure 1 shows the photographs of a back assistive device equipped with the resistant force canceller (RFC) that we developed. The device can be worn with shoulder straps and waist belt

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Fig. 1 Back assistive device equipped with a resistant force canceller (RFC). (a) swing arm, (b) rubber belt, (c) non-elastic lower belt, (d) plastic frame, (e) mesh, (f) rubber band to pull up the swing arm and remove slack of lower belt assembly.

like a backpack. A lower belt assembly is attached to the swing arm mounted on the frame. A rubber belt is placed in series with the lower belt. The belt is stretched and produces an assistive moment using the plastic frame as a lever in a forward bending motion. As fulcrum of the lever, a mesh is attached to the frame to decrease the contact pressure between the frame and the back.

The main part of the RFC is composed of two plastic arms produced by a 3D printer. Each arm is connected to the frame by a pin near the center of the frame. A belt is connected to the other end of the arm so as to rotate in the same direction while walking. Under walking conditions, the lower belt of the leading leg is pulled downward. On the contrary, the lower belt of the trailing leg is slackened. Then, each arm rotates in the same direction without any difficulty due to the configuration of the arm. The rocking motion of the two arms while walking allows almost resistance-free walking. However, when the person is bending forward, both lower belts are pulled downward simultaneously, which would result in both arms rotating in opposite directions. However, they cannot rotate in the opposite direction due to the configuration of the arms. Thus, the rubber belts are stretched during the forward bending motion. The tension of the rubber belts is transformed to an assistive moment by the frame. As a result, the efforts of back musculature are reduced. This simple mechanism facilitates walking with a back support device without reducing the effectiveness of back support.

The effectiveness of the RFC mechanism was evaluated by measuring the belt tension under walking conditions as shown in **Fig. 2**. The measuring apparatus was carried on a cart to keep a constant distance from the walking subject. The belt tension was measured using a strain gauge as shown in **Fig. 2**. Prior to the experiment, the length of each lower belt was adjusted to eliminate any sag in the standing posture. The subject was a young man 22 years of age, 1.75 m in height and 68 kg in weight. Step length was maintained at 60 cm.

Before commencing the experiment, the aim and protocol of the experiment were explained to the subject. The subject read the written explanations and gave informed consent. Anonymity was

Fig. 2 Lower belt tension under walking condition evaluated by a strain gauge with a cart carrying measuring apparatus.

Fig. 3 The lower belt is stretched by flexion of the hip joint.

maintained for the protection of personal information.

3. Theory

3.1 Change of lower belt length in gait

Figure 3 shows a simplified mechanical model of a leg with a lower belt in the sagittal plane. O_H denotes the center of hip joint. Point B denotes the lower edge of the back assistive device. L_0 denotes the length of the lower belt in a standing posture. L_θ denotes the length of the lower belt with the hip joint at a flexion angle θ_H . Because the leg and the lower belt have different centers of rotation, the length of the lower belt L_θ increases with increase in flexion angle θ_H . The increment of the belt length is denoted by δ . From a geometrical consideration,

$$L_0 = \sqrt{(H - h)^2 + b^2} \quad (1)$$

$$L_\theta = \sqrt{O_H B^2 + H^2 - 2O_H B \cos(\phi + \theta_H)} \quad (2)$$

$$\phi = \tan^{-1}(b/h) \quad (3)$$

$$\delta = L_\theta - L_0 \quad (4)$$

Provided $H = 800$ mm, $b = 130$ mm and $\phi = 77^\circ$, the increment of

Fig. 4 The length of lower belt increases with increase in flexion angle of the hip joint, causing the swing arm to rotate.

Fig. 5 Mechanical model of RFC.
(a) in standing posture, (b) under walking condition.

lower belt length is calculated using Eqs. (1) to (4). **Figure 4** shows the results. According to these results, the increment of the lower belt length is 68 mm at the flexion angle of the hip joint $\theta_H = 30$ degrees. The flexion angle of the hip joint at a typical walking speed is approximately 30 degrees [9]. Therefore, as far as this model is concerned, an extra belt length of 68 mm is required to achieve stress-free walking with the back assistive device.

3.2 Mechanism of RFC

Figure 5a shows a mechanical model of the RFC used in this study, in a standing upright posture. P_A and P_B denote belt tension. A and B denote symmetrically placed swing arms, rotating around points O_1 and O_4 , respectively. They are in contact at C located on an arc with radius R . Since the distance between O_1

and O_4 and the distance between O_2 and O_3 are the same, this mechanism is considered to be a parallelogram [10] of four-bar linkage, as far as the two swing arms are in contact at C. Thus, they can rotate in the same direction for the same rotational angle θ . **Figure 5b** shows the states of the swing arms after rotating through angle θ in a counterclockwise direction. Let $\mathbf{r}_{AC}$ be the vector from O_1 to C.

$$\mathbf{r}_{AC} = (h \sin \theta - R)\mathbf{i} - h \cos \theta \mathbf{j} \quad (5)$$

Let $\mathbf{r}_{E0}$ be the vector from O_1 to E in the state of $\theta = 0$.

$$\mathbf{r}_{E0} = (L - R)\mathbf{i} + H\mathbf{j} \quad (6)$$

Let $\mathbf{Q}$ be the rotation matrix of a vector around z axis [11].

$$\mathbf{Q} = \begin{pmatrix} \cos \theta & -\sin \theta \\ \sin \theta & \cos \theta \end{pmatrix} \quad (7)$$

The angle of rotation is θ in the sense of a right-handed screw travelling in the positive z direction. The vector $\mathbf{r}_E$ in the state of **Fig. 5b** is given by

$$\mathbf{r}_E = \mathbf{Q} \cdot \mathbf{r}_{E0} \quad (8)$$

Let $\mathbf{R}_C$ be the force exerted by the mating arm through the contact point C. Neglecting the frictional force, equilibrium of moment about the point O_1 is

$$\mathbf{r}_E \times \mathbf{P}_A + \mathbf{r}_{AC} \times \mathbf{R}_C = \mathbf{0} \quad (9)$$

Substituting Eqs. (5) and (8) into Eq. (9) gives

$$\mathbf{R}_C = \frac{(L - R) \cos \theta - H \sin \theta}{h \cos \theta} \mathbf{P}_A \quad (10)$$

In the same manner, equilibrium of moment of the swing arm B can also be obtained. Let $\mathbf{r}_{BC}$ be the vector from O_4 to C.

$$\mathbf{r}_{BC} = (h \sin \theta + R)\mathbf{i} - h \cos \theta \mathbf{j} \quad (11)$$

Let $\mathbf{r}_{F0}$ be the vector from O_4 to F in the state of $\theta = 0$.

$$\mathbf{r}_{F0} = (R - L)\mathbf{i} + H\mathbf{j} \quad (12)$$

Let $\mathbf{r}_F$ be the vector from O_4 to F in **Fig. 5b**.

$$\mathbf{r}_F = \mathbf{Q} \cdot \mathbf{r}_{F0} \quad (13)$$

Equilibrium of moment about the point O_4 is

$$\mathbf{r}_F \times \mathbf{P}_B + \mathbf{r}_{BC} \times \mathbf{R}_C = \mathbf{0} \quad (14)$$

Substituting Eqs. (11) and (13) into Eq. (14) gives

$$\mathbf{R}_C = \frac{(L - R) \cos \theta + H \sin \theta}{h \cos \theta} \mathbf{P}_B \quad (15)$$

Using Eqs. (10) and (15), the contact force $\mathbf{R}_C$ can be eliminated and the ratio of belt tension is derived.

$$\frac{P_B}{P_A} = \frac{(L - R) \cos \theta - H \sin \theta}{(L - R) \cos \theta + H \sin \theta} \quad (16)$$

In the case of symmetric forward bending of the trunk, the ratio of the belt tension becomes $P_B/P_A = 1$ and angle $\theta = 0$. However, in the case of asymmetric bending, the swing arm may rotate for an angle θ so as to satisfy Eq. (16).

The extensible length of the lower belt with rotation of arm B can be calculated by considering the displacement vector $\mathbf{u}_F$ of point F from $\mathbf{r}_{F0}$ to $\mathbf{r}_F$.

$$\mathbf{u}_F = \mathbf{r}_F - \mathbf{r}_{F0} \quad (17)$$

Substituting Eqs. (12) and (13) into Eq. (17) gives

$$\mathbf{u}_F = \begin{pmatrix} (L - R)(1 - \cos \theta) - H \sin \theta \\ -(L - R) \sin \theta - H(1 - \cos \theta) \end{pmatrix} \quad (18)$$

In the same manner, the displacement vector $\mathbf{u}_E$ is given by

$$\mathbf{u}_E = \begin{pmatrix} -(L - R)(1 - \cos \theta) - H \sin \theta \\ (L - R) \sin \theta - H(1 - \cos \theta) \end{pmatrix} \quad (19)$$

The dimensions of the device are: $L = 81$ mm, $H = 17.6$ mm and $R = 15$ mm. Substituting these values into Eqs. (18) and (19), the displacement of points F and E at angle θ can be calculated. For simplicity, equating an increment of the lower belt length δ to the magnitude of vertical component of the displacement vector $\mathbf{u}_F$,

the relation between the flexion angle of hip joint θ_H of the model and the corresponding rotation angle of the swing arm θ can be calculated. The result is shown in **Fig. 4**. It can be seen that the relation of $\theta \approx 2\theta_H$ and the increase in lower belt length δ , which is caused by walking, is absorbed through rotation of the swing arm.

The sliding velocity at the contact point C can be calculated. Let ω be the angular velocity of the swing arm, $\mathbf{V}_{AC}$ be the velocity of the point C of arm A, and $\mathbf{V}_{BC}$ be the corresponding velocity of arm B.

$$\mathbf{V}_{AC} = \omega \times \mathbf{r}_{AC} \quad (20)$$

$$\mathbf{V}_{BC} = \omega \times \mathbf{r}_{BC} \quad (21)$$

Because the profile in the vicinity of contact point C is an arc with radius R, the common unit tangent vector at C is the base vector $\mathbf{j}$.

Making use of Eqs. (5), (11), (20) and (21), the sliding velocity s can be calculated.

$$s = (\mathbf{V}_{BC} - \mathbf{V}_{AC}) \cdot \mathbf{j} = 2\omega R \quad (22)$$

4. Results

Figures 6 show the changes in belt tension with time under walking conditions. **Figure 6a** shows the changes in belt tension without a RFC mechanism. **Figure 6b** shows the changes with a device employed in the previous study [8]. **Figure 6c** shows the changes using the device developed in this study. It is evident that the RFC significantly reduced the belt tension under walking condition. Using the swing arm RFC, the peak belt tension was reduced to approximately 23% of the tension without an RFC mechanism. The peak value decreased to one-half of that when a former slide string type RFC was used.

5. Discussion

The smallest belt tension was obtained when the swing arm mechanism was used. When the slide string mechanism was used, small peaks were observed even in the trailing leg as shown in **Fig. 6b**, because the leading leg and the trailing leg were connected by a string. When the swing arm mechanism was used, small peaks in the trailing leg disappeared as shown in **Fig. 6c**. However, the belt tension peaks did not disappear completely, which may be due to the small radius of the swing arm. Experiments showed that the belt tension peaks disappeared when longer arms were used. However, the total width of the arms became larger than the width of the frame. The dimension of the swing arm was decided by the dimension of the plastic frame that had been determined by an industrial designer. As far as this frame is used, it is difficult to completely eliminate the belt tension peaks. Since the device provides adequate assistive moment for the back, the presence of slight belt tension is not considered to be a serious problem when using this device.

6. Conclusion

Change in lower belt length in gait was examined using a model. A RFC was devised based on the reciprocal motions of the legs under walking conditions. This device consists of two matching swing arms oscillating synchronously with walking movement. The swing arms rotate in the same direction while walking, thereby cancelling the resistant force generated from the elastic element caused by walking. On the other hand, in a bending forward posture, the swing arms are locked and an assistive moment for the back is generated from the elastic elements. The mechanism was analyzed using a simple model and its effectiveness was confirmed by an experiment. This mechanism was effective in reducing undesirable tension of the lower belt caused by walking. Using the RFC mechanism, the magnitude of belt tension was reduced to approximately 23% of that without an RFC mechanism, while the back supporting property of the device was maintained.

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Fig. 6 Changes in belt tensions with time under walking condition without RFC mechanism (a), with a slide string type mechanism (b), and with a swing arm type mechanism (c).

References

1. Zhuang Z, Stobbe TJ, Hsiao H, Collins JW, Hobbs GR: Biomechanical evaluation of assistive devices for transferring residents. *Appl Ergon.* **30**, pp. 285–294, 1999.
2. Abdoli-E M, Agnew MJ, Stevenson JM: An on-body personal lift augmentation device (PLAD) reduces EMG amplitude of erector spinae during lifting tasks. *Clin Biomech.* **21**, pp. 456–465, 2006.
3. Abdoli-E M, Stevenson JM: The effect of on-body lift assistive device on the lumbar 3D dynamic moments and EMG during asymmetric freestyle lifting. *Clin Biomech.* **23**, pp. 372–380, 2008.
4. Lotz CA, Agnew MJ, Godwin AA, Stevenson JM: The effect of an on-body personal lift assist device (PLAD) on fatigue during a repetitive lifting task. *J Electromyogr Kinesiol.* **19**, pp. 331–340, 2009.
5. Frost DM, Abdoli-E M, Stevenson JM: PLAD(personal lift assistive device) stiffness affects the lumbar flexion/extension moment and the posterior chain EMG during symmetrical lifting tasks. *J Electromyogr Kinesiol.* **19**, e403–e412, 2009.
6. Whitfield BH, Costigan PA, Stevenson JM, Smallman CL: Effect of an on-body ergonomic aid on oxygen consumption during a repetitive lifting task. *Int J Ind Ergon.* **44**, pp. 39–44, 2014.
7. Verbeek JH, Martimo KP, Kuijer PPFM, Karppinen J, Viikari-Juntura E and Takala EP: Proper manual handling techniques to prevent low back pain, a Cochrane Systematic Review. *Work.* **41**, pp. 2299–2301, 2012.
8. Gushikata A, Shigemoto T, Kozen I, Miura A, Imado K: Development of simple back assistive device. *Proc of WACBE (5th world congress on bioengineering)* Taiwan, Tainan, pp. 18–21, 2011.
9. Neumann DA: *Kinesiology of the Musculoskeletal system*, Mosby, pp. 536–537, 2002.
10. Norton RL: *Design of Machinery*, 2nd ed. WCB/McGraw-Hill, USA, p. 49, 1999.
11. Spencer AJM: *Continuum mechanics*, Longman, London, p. 64, 1985.

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