공학석사학위논문

리프팅에서의 근력 보조와 자세 가이드를 위한 유연한 착용형 장치 개발

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Abstract

Development of Soft Wearable Device for Muscular Assist and Posture Guide at Lifting

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There is a saying: comfortable posture is usually bad posture. A human feels comfortable and consumes less energy when holding the object on the floor by flexing his back than bending leg, because of characteristics of human body design. However, from a spinal point of view, lifting a heavy load in a back bent posture is a dangerous act that may increase the risk of a disc injury by 2 to 4 times. We have developed a soft wearable device that can reconfigure these contradictory human body systems to make to feel the correct posture comfortable and bad posture uncomfortable. A representative posture correction functions of this device is to correct the wearer’ s habit to use a lifting technique called squat instead of the stoop and to induce wearer to keep his knee behind the toes in the process of bending the legs. Experiments with varying lift posture showed that this device interfered with the stoop motion but assisted the squat motion. Also, we proved that the muscular force assistive effect of the device increased in knee backed posture. Another experiment was conducted in which the 15kg box was repeatedly lifted at a rate of 10lifts/min for 6minutes. In case of taking squat posture, EMG values of spinal erector, biceps femoris, and rectus femoris decreased by 14.0%, 17.0%, and 14.3% respectively when wearing the device. Also, in this case, the metabolic energy cost decreased by 29.5% and the heart rate decreased by 13.6%, proving that the device is effective in reducing fatigue of the worker. Compared to stooping without a device, the result was a 24.7% reduction in metabolic energy cost and a 7.3% reduction in heart rate when the subject takes squat posture with wearing the device. Therefore, the device is expected to help stuff transportation workers to lift objects in correct posture with small fatigue.

Keyword : Soft robot, Wearable robot, Lifting, Posture correction, Muscular force support, Metabolic cost

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Chapter 1. Introduction

1.1. Study Background

Work-related Musculoskeletal Disorders (WMSDs) are serious issue at industries. In USA, totally 356,910 cases of WMSDs occurred in 2015. This accounts for 31% of total nonfatal injuries of workers [1]. WMSDs cause labor time loss and compensation cost expenditure for workers. Also, it sometimes leaves a sequela to the workers that cannot be recovered. The Institute in Medicine estimates $45-54 billion economic loss occurs annually in USA because of WMSDs [2, 3]. Especially, a heavy lift is an act that causes many injuries, equivalent to 30% of cases of total WMSDs [1]. Therefore, a solution that can protect workers when they lift heavy objects is required.

Many wearable robots and devices to assist lift have been developed so far. The most common type has rigid skeleton with joint and uses a motor as an actuator [4-7]. Pneumatic actuators are also often used as alternatives to motors [8-10]. They reduce work fatigue by supporting erector muscle strength with actuator when raising upper body. Some devices with rigid skeleton use energy storage element like a spring [11-13]. They store energy in the spring at lowering process and emit it at lifting. Also, there are soft type wearable devices that do not use rigid skeleton [14-21]. They are made of soft materials like fabric or cable and partially contain rigid elements. Most of them use elastomer or spring as a power source[16-21], but some use active actuator[14, 15].

All of these devices have an effect on reducing the risk of muscle injuries through muscular force support in lifting. However, their design is biased only to provide a lot of force, and they less consider lifting posture of a wearer. Because of shape of human spine, it is inevitable that the load is concentrated on the front of the disc when flexing the back and possibility of herniation becomes high [22−25]. Therefore, there is a limitation in preventing lift in−jury through only the muscular force support without correcting the posture.

Taking correct posture is important to prevent injury in heavy lift. Lifting a heavy object in a bent back posture is a dangerous be−havior, whose relative risk is 2 to 4 [26]. This means that the risk of injury can increase up to 400%. However, just by taking correct posture at lift can reduce the relative risk to 0.71. As a safe lifting posture, industrial health care corporations and personal trainers recommend to keep the back straight and the knee behind toes: this motion is called 'proper squat from' or 'semi−squat form', which is similar with deadlift exercise [27−30].

1.2. Purpose of Research

Until now, wearable robot or device researchers have consid−ered that it is important to realize all the degrees of freedom (DOF) of human being while supporting enough force. In order to realize this, they analyzed the human body system and designed the wear−able devices in such a way as to mimic the arrangement and shape of human joints, muscles, and other elements.

However, human body system is designed to use less energy and to feel comfortable when stooping than squatting [31−33]. Therefore, if people do not pay attention, they unconsciously lift objects by back flexing posture and exposed to injury. Therefore, mimicking the human body system is not helpful in changing one's lifting habit and inducing people to use correct lifting techniques. To induce wearer to correct lifting posture, adequate interruption is required on an improper motion such as back flexion. Therefore, we developed a new device design to change human body system to feel comfortable at correct posture and to feel uncomfortable at bad posture.

In this study, we introduce a soft wearable device that induce people to follow correct lifting technique with small fatigue by changing human body system. For that, our device contains two representative functions: motion guide and muscular force support. Novel approach for developing a soft wearable device for motion guide will be suggested. Then, a discussion about design strategy of device required to implement functions will be followed with the introduction of our anchoring and routing technologies. Also, an evaluation about the effectiveness of posture guiding and muscular force supporting functions were conducted by experiments. Finally, this paper will be concluded with summary and discussion.

Chapter 2. Design

A developed soft wearable device is shown in figure 2.1. It is a passive device that does not have electrical elements. In general, passive wearable devices are mainly composed of an elastomer. However, since the elastomer cannot influence the positional relationship between points, the elastic cable is not suitable for motion limiting, or posture guiding. Therefore, our device is developed using an inelastic cable (webbing strap) and elastic elements are only used partially. This new approach has caused new issues that were not addressed in previous studies. First, how the device can distin- guish between proper and improper motion? And how to implement the degree of the assistance or interruption differently according to the posture? Second, how to provide muscular force support with inelastic cable? The solutions for these issues will be presented in the following sessions.

Fig. 2.1. Prototype of soft wearable device for lift assist

2.1. Design Strategy

Human motion is manifested by the combination of rotation of joints. It means that the joints must rotate along a certain angular relationship in order to make a certain intended motion. In the original human system, the spine, hip, and knee joints do not have a particular relationship, so they can move independently. Therefore, the number of motions that can be taken during the lowering and lifting process is infinitely large, and people prefer the least energy-intensive motion (stoop posture). We must limit the rest of the joint angle combinations except for inducing the correct posture among the many combinations of joint movements that a person can take. For this, spine, hip, and knee joint angles have to be coupled as the following relationship: hip joint can rotate only when the knee is rotating, and spine always has to maintain the neutral angle. These can be implemented if the joints are properly connected each other by the inelastic cable and the cable passes through the appropriate path.

The most important element in implementing the motion guide and muscular support function is the cable releasing mechanism used at the knee. Most lift assisting soft wearable devices attaches a strap at the thigh (Fig. 2.2.(a)) or just below of the knee (Fig. 2.2.(b)) to fix their device [16-21]. Because these designs do not couple the hip and knee joint, when these designs are applied to an inelastic cable, the motion of the hip joint is always limited regardless of the state of the knee. Another simple design can be thought as a solution of this issue. If the cable passes behind the knee like figure 2.2.(c), knee bending causes loosening of the cable. Consequentially, the cable does not interrupt lowering motion if knee and hip move at the same time. We expected this design can guide squat motion to a wearer, but because the too much cable is released at the knee, the device could not block back flexion even when the knee is slightly bent. Also, the device could not support muscular force at squat posture because the cable is loosened. As a solution, we developed a novel cable connection method, named cable releasing mechanism (Fig. 2.3). Cable releasing mechanism reduces cable releasing length at the knee to half of the previous design' s (Fig. 2.2.(c)), and allows the cable remains tight in the squat posture, which is essential for posture correction and muscular force support.

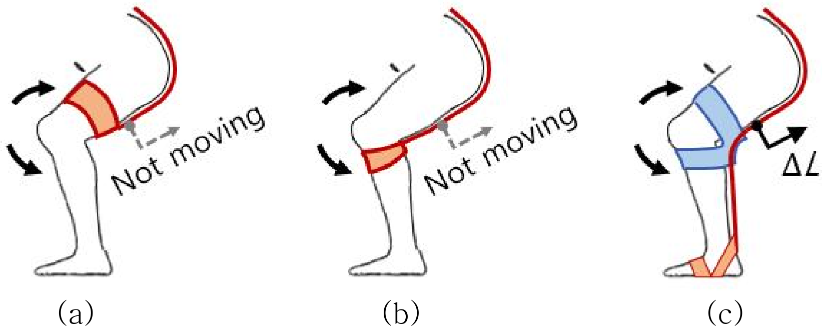


Fig. 2.2. Various available leg anchor designs. (a) Fixing at thigh. (b) Fixing at just below of the knee. (c) Simple design that cable passes behind the knee.

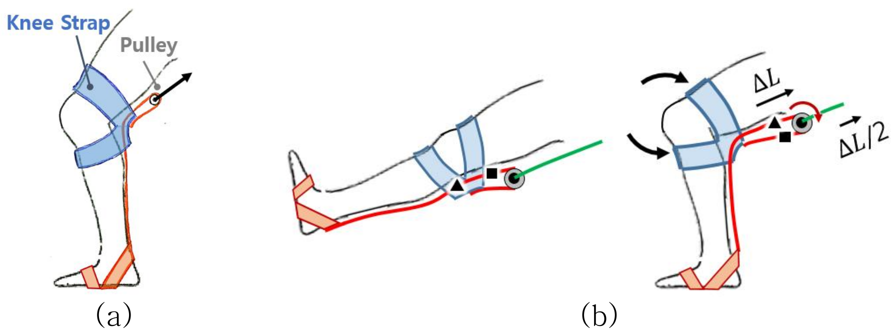


Fig. 2.3. Cable releasing mechanism. (a) Developed leg anchor design. (b) Mechanism working principle.

2.1.1 Posture Correction Principle

The goal is to change the wearer's lifting habit to use the squat lifting technique, while helping to keep the back in neutral position while the lift process. Also, it has to encourage the wearer to keep the position of the knee as much as possible while squat movement, not to protrude than the toes. In case of a back, it is sufficient to maintain the neutral position by pulling the shoulder to backside with inelastic cable when the wearer tries to lower his body. However, in the case of a knee, it should be possible to take various postures while wearing the device for smooth transportation work. Therefore, we should not approach in a way that forcefully pulling the knee backward, but rather inducing the wearers to choose the correct knee posture themselves. For this purpose, we designed the device so that when the wearer pulls the knee to the backside, a strong assist force is generated, and as the knee is protruded forward, the assist degree becomes smaller.

As the cable passes over the surface of the hip, the cable is pulled when people bend the hip joint. The cable releasing mechanism allows the cable to loosen in an appropriate amount when the knee is bent. When people lower their body while keeping the cable tight, the length pulled from the hip (ΔL_{hip}) and length released from the knee (ΔL_{knee}) are kept the same according to the constant length condition of the cable. ΔL_{hip} and ΔL_{knee} are functions of the angle of the hip (θ_{hip}) and the knee (θ_{knee}) and determined by cable path, anchor strap shape and wearer's body shape. Therefore, guided motion trajectory can be tuned by changing design parameters of the device. We analyzed ΔL_{hip} and ΔL_{knee} functions by an experiment with video marker tracking (Fig. 2.4). Simple models to show the action of the device in the knee and the hip are shown in figure 2.5 and equations are derived as follows.

$$\Delta L_{knee} = f_{knee} R_{knee} sin\frac{\theta_{knee}}{2} \quad (1)$$

$$\Delta L_{hip} = f_{hip} R_{hip} \theta_{hip} \quad (2)$$

Where R_{knee}=60mm and R_{hip}=120mm, which are the radius of the knee joint and the hip joint.

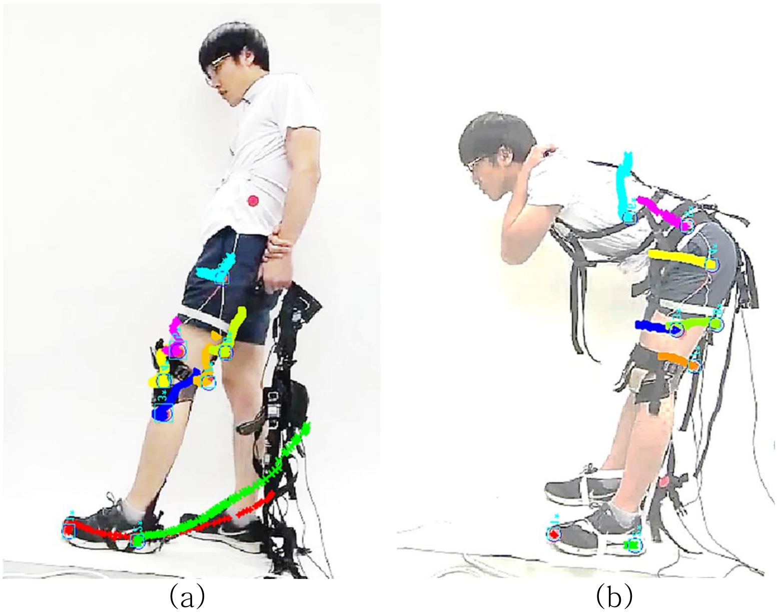


Fig. 2.4. Marker position and tracking result. (a) Knee flexion. (b) Hip flexion.

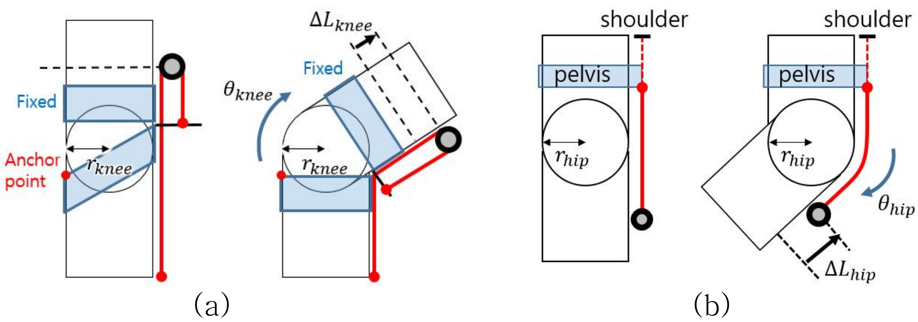
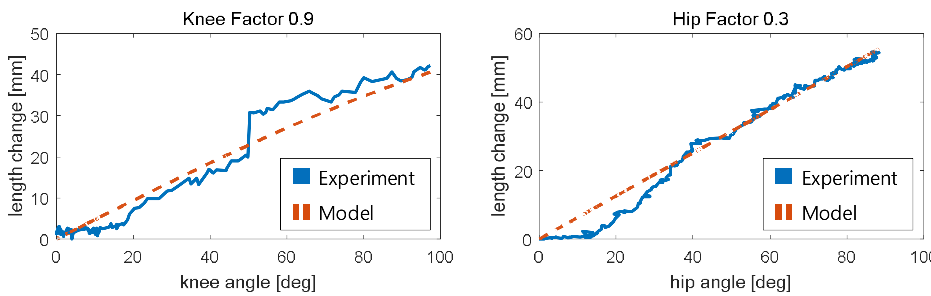


Fig. 2.5. Mathematical model for the cable movement at each joint. (a) Knee joint. (b) Hip joint.

Because there is no exact human body model that can reflect stiffness of muscle and skin, the modeling was conducted by multiplying unknown factors (f_{knee}, f_{hip}) to the model, carried from rigid body assumption. These factors are expected to contain information about the softness of the human body. We found proper values for these factors by matching experiment and modeling results: 0.9 for f_{knee} and 0.3 for f_{hip} (Fig. 2.6). By applying constant length condition of cable, we can get theoretical angle relationship equation of hip and knee joints as follows (Fig. 2.7).

$$0.9R_{knee}sin\frac{\theta_{knee}}{2} = 0.3R_{hip}\theta_{hip} \quad (3)$$



(a) (b)

Fig. 2.6. Cable length change plots of experiment and model. (a) Length change at the knee (f_{knee}=0.9). (b) Length change at the hip (f_{hip}=0.3).

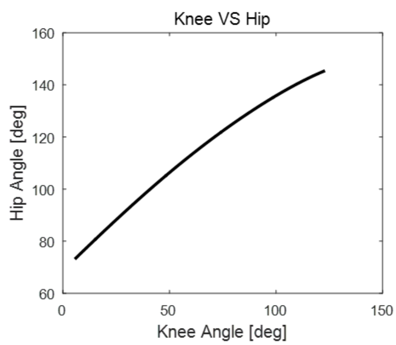


Fig. 2.7. Theoretical angle relationship between knee and hip joints.

In the real situation, however, a relationship of hip and knee joints in equation (3) is not strictly followed because of compliance, which comes from soft elements of human body and the elastic component of the wearable device. However, the wearable device still induces the wearer to act within a range that is close to the aimed motion represented in (3). If the wearer pushes the knees further forward than intended in (3), the knee angle increases and the cable loses its tension (Fig. 2.8). In this case, since the assist force transmitted to the wearer by the device becomes smaller, the wearer becomes more fatigued than when taking correct posture. As the knee is pulled backward, the cable becomes taut and the wearer can receive a lot of assists. However, if wearer tries to bend the hip joint too much or unbend the knee joint so that the posture is too much different from (3), the movement is limited by the tension of the cable. Figure 2.9 explains the posture correction process.

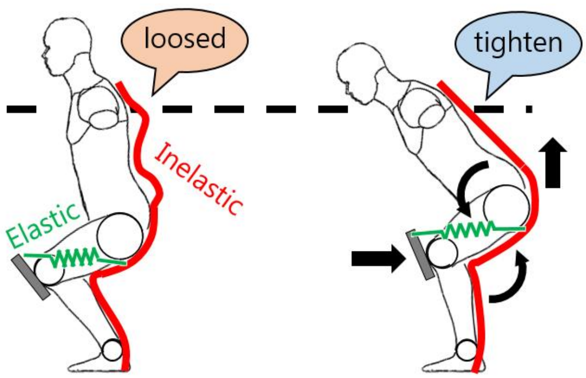


Fig. 2.8. Illustration showing the relationship between knee posi- tion and cable tension.

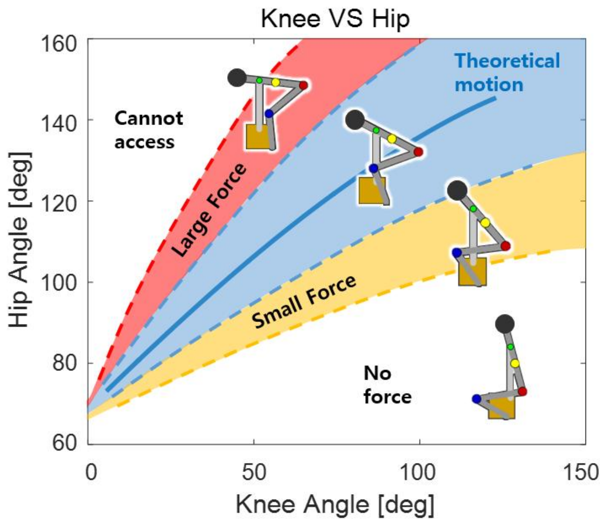


Fig. 2.9. Illustration explaining the posture correction process.

2.1.2 Muscular Force Support Principle

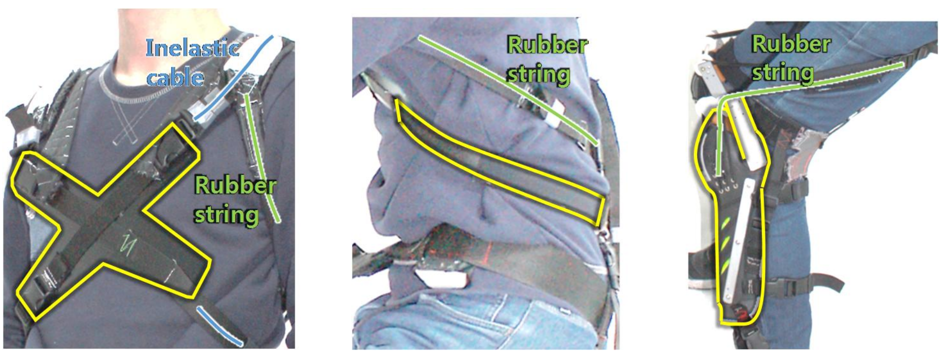
In general, passive wearable devices generate force by energy storage and emission. How can we get an assistive effect like this with an inelastic cable? First of all, keep in mind that since the muscle and skin are made of soft tissue, pulling the body with an inelastic cable will show a spring-like force profile for a while (Of course, it is very stiff and has low strain). When the wearer tries to stoop, the cable is tightened by the extension of the length by the bending the back and rotation of the hip joint. Because of the use of inelastic cable, the tension increases very sharply (refer to experiment result in figure. 18-red line). In this case, tension acts to limit movement. On the other hand, when the wearer tries to squat, the cable is pulled at the hip joint but released at the knee. This results in the effect that the cable tightens slowly while wearer lowering the posture (refer to experiment result in figure. 18-blue line). In this case, the tension acts like a spring to help the hip joint motion and helps to maintain the neutral state of the back.

The back and hip joint became able to receive sufficient muscular assist, but in the case of the knee, the tension of the cable contributed to the knee flexion, which caused the problem of increasing the muscle power consumption during the lift operation. To solve this problem, it is necessary to reduce the tension of the cable passing behind the knee and to apply additional force to contribute to the knee extension. We found that the proper placement of elastic elements in this device can solve this problem, which will be introduced later (section 2.2.1). Furthermore, since the elastic element does not limit the length, the posture correction strategy mentioned in the previous chapter is still valid.

2.2. Anchor Design

The anchoring means fixing the device to the body so that it does not come off the body when force is applied, and creating an environment that can use the desired area as a point of force. Since it is not possible to attach a cable directly to a human's skeleton, finding an alternative way of fixing it is an issue. Especially in passive devices, anchoring is very important because the anchoring performance determines the magnitude of the device force. Also, anchor design determines the position and direction of the force acting on the wearer's body from the wearable device. Since the human body is a very complex linkage system, the posture inducing performance strongly depends on the position and direction of the force applied.

Existing soft wearable devices for lift assist have developed their own anchoring technology. They are slightly different in shape but commonly hang a loop on the shoulder and pull it down, which is similar to a strap used on the backpack [19-21]. When using these anchor, the force is transmitted from the wearable device to the wearer in the form of symmetrically pressing the front and back of the shoulder. These ways can generate torque for lumbar spines, including L4-L5 discs and can reduce trunk muscle fatigue. However, in order to maintain straight back posture regardless of the wearer's intention, it is necessary to apply forces that can generate torque for entire trunk including thoracic spine, shoulder as well as a lumbar spine. To do that, the device has to be designed to apply a force an asymmetrical form of front and back based on the shoulder. When using an asymmetrical anchor design, a new issue occurs in which the device slides on the body as it rotates about the shoulder. Our anchor also must be able to solve this issue. In following sub-sessions, our anchoring technologies that used to solve each issue will be introduced.



(a) (b) (c)

Fig. 2.10. Anchoring parts of the wearable device. (a) X-shaped plate. (b) Sloped trunk strap. (c) Knee lever arm structure.

2.2.1 Anchoring Issue#1: Force & Torque Setting

As mentioned before, the anchor should be designed so that the force and torque can be used for strength assistance and posture correction during squat lifting. Figure 2.10 shows anchoring parts of our wearable device and figure 2.11 shows schematic of device wearer and action of force and torque during the lift. The X-shaped plate pushes the chest back and induces a cable to push the shoulder back when tension is generated (Fig. 2.10.(a)). The X-shaped plate is connected to the strap that surrounds the trunk (Fig. 2.10.(b)). As this strap passes over the lumbar spine of the back, it pushes the lumbar part forward when tension is generated. These two elements help the wearer to maintain the S shape of the entire spine by pushing the shoulder and thoracic spine backward and lumbar spine forward during the lifting (Fig. 2.11).

Because the cable passes through the backside of the knee, when the cable becomes taut, the knee receives a negative torque in the direction that the flexion is induced. To solve this problem, it is necessary to reduce the negative torque component and to increase the positive torque component by moving a part of the force behind the knee to the front of the knee. However, there is a problem that the knee does not have sufficient moment arm. There is also an issue that the front of the knee is not proper to be used as a cable path because exposed ligaments at the front of the knee are vulnerable to external stimuli. We designed lever arm structure to solve these problems (Fig. 2.10.(c)). It has a bar extending from the knee joint, with the end of the bar is connected to the cable by rubber string as shown in figure 2.11. When the leg is bent, the tension of the cable passing behind the knee becomes smaller, but the tension of rubber string becomes stronger. Therefore, the lever arm structure reduces the negative torque on the knee while producing positive torque. The rigid frame of the lever arm structure also acts to transfer the force to the skeleton while avoiding the ligaments when the knee is subjected to force of rubber string.

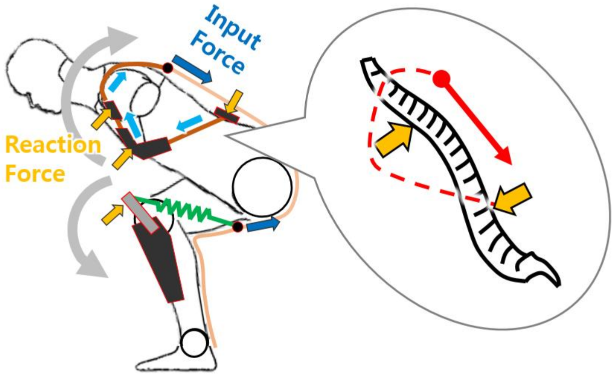


Fig. 2.11. Schematic of force and torque generated by anchoring parts.

2.2.2 Anchoring Issue#2: Fixing

Because most conventional devices use a backpack strap at the shoulder, fixing the device is not that difficult. However, in our device, the cable is held in the torso and the shoulders are used only as pulleys, which causes the issue that the rotation of the device should be prevented. In addition, there is a difficulty in designing the anchor to prevent the cable from passing through the area where the nerve is concentrated. For example, if we design an anchor in the form of a cable wrapped around wearer's armpit, the anchor can get good fixing performance, but we had to give up this design because the nerves are concentrated in the armpit. The X-shaped plate in figure 2.10.(a) plays an important role in firmly fixing the device while avoiding underarm compression. The X−shaped plate can be thought of as dividing the upper(∨) and lower(∧) parts. The upper(∨) part maintains the distance that the cable does not push the around of the armpit, and at the same time it induces cable to pass the front of the shoulders to push them backside. The lower(∧) part serves to fix the plate so that it does not go up when the cable is pulled. The key to fixing the device to the body with the X−shaped plate is to use a sloped strap at trunk as shown in figure 2.10.(b). If a non−tilted horizontal strap is used, it always has to be tightened to prevent the device from being pulled up. However, in our anchor design, it is possible to make the strap loosen in the standing state and tighten automatically only when squatting and prevent slipping (Fig. 2.12.(a)). Also, it can be seen that the tilted form is advantageous when considering the direction of the force generated on the cable. Since the cable cannot withstand the shear direction and can only withstand the tensile direction, a certain amount of slip is inevitable in the case of a horizontal strap (Fig. 2.12.(b)). On the other hand, a sloped strap can resist vertical forces without slipping (Fig. 2.12.(c)).

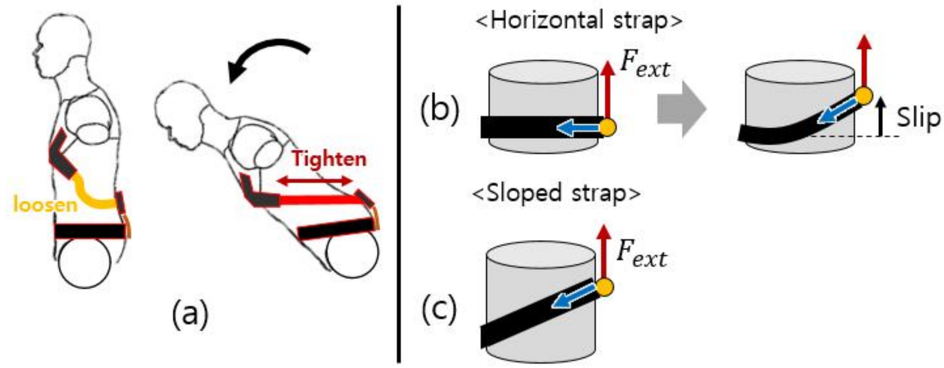


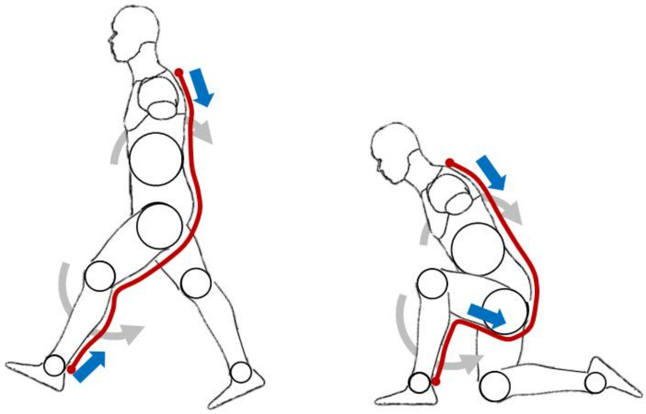
Fig. 2.12. Advantages of using sloped trunk strap. (a) Tension changed according to posture. (b) Slipping issue of horizontal strap. (c) Slipping issue solved by using sloped strap.

2.3. Cable Routing

Cable routing means to determine what path a cable will go through and to create an environment that cable can move con-stantly along a given path. Since the torque value and ratio applied to the joints vary with the distance between each joint and cable path, it is important to design the cable path well for proper posture induction and force assist. In particular, in our device, an inelastic cable uses a hip as a pulley as it passes through the hip surface. Because the surface of the hip is curved, the degree to which cable is pulled when bending the hip joint depends on cable path. And since the inelastic cable plays a role in restricting the movement of the joints, it is necessary to design the cable path well so that affect to joints related with incorrect lifting motion and not affect to other joints. In following sub-sessions, our cable routing technologies that used to solve each issue will be introduced.

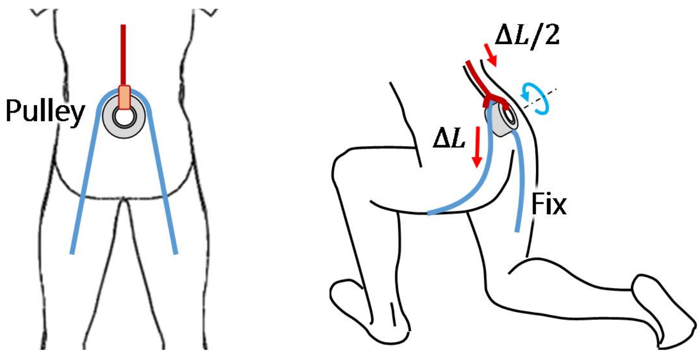
2.3.1 Routing Issue#1: Enabling Leg Motions

This wearable device induces the correct lift motion by coupling the leg joints and the spine. However, even when the leg is used for a purpose other than the lift, the leg motion is restricted and the back is pulled. Typically, walking and kneeling movements are interrupted, so it is difficult to take stuff transportation work (Fig. 2.13). Since these two movements occur when both legs intersect, these two movements have a commonality in that only one of the two hip joints has a large angle. Using this feature, we developed a cable path that generates a differential motion using a pulley as shown in figure 2.14 to prevent motion interruption. Originally, the cable was pulled the same amount when bending both legs and one leg, but using this method can reduce interference to the movement because it is only pulled by half when bending one leg. As a result of the test, the application of this cable path reduced the disturbance to the walking motion and made the kneeling motion easier.



(a) (b)

Fig. 2.13. Leg motions that can be restricted by the device. (a) Walking motion. (b) Kneeling motion.



(a) (b)

Fig. 2.14. Differential mechanism for soft wearable device. (a) Cable path for differential mechanism. (b) Working of the differential mechanism at kneeling motion.

2.3.2 Routing Issue#2: Maintaining the Cable Path

In our device, the cable is tethered loosely from the shoulder to the foot, because there should be no pre-tension in neutral standing posture. Therefore, there is a possibility that when the external force is applied, the strap may shake or change its path (Fig. 2.15.(a)). Also, because the cable goes past the hip surface, which is a curved surface when the straps are tensed as the wearer bend legs, cable tends to slip down to the small radius area along the surface of the hip (Fig. 2.15.(b)). This not only makes the fit worse but also reduces the performance of the device because the bend of the hip joint does not pull the cable as intended. To solve these issues, there must be body-mounted elements in the space between the ends of the long cable, and the path must be designed so that the cable can withstand the horizontal force. Figure 2.16.(a) shows our novel cable path, suitable for use on wearable devices using cables. By attaching a strap that surrounds the pelvis, a mount point for the body was secured. Using this, we constructed a pulley system as shown in figure 2.16.(a). The horizontal position of the pulley at the leg can be fixed without moving because the cable is tilted obliquely to the left and right with respect to the pulley (Fig. 2.16.(b)). It is also an issue to adjust the degree to which the cable is pulled from the hip joint. This design has the advantage that it is easier to control the path of the cable to the hip surface by changing cable length or angle. In addition, using this cable path, the tension on the shoulder is about half of the tension on the leg, so it can apply a great deal of force to the leg without putting a strain on the spine.

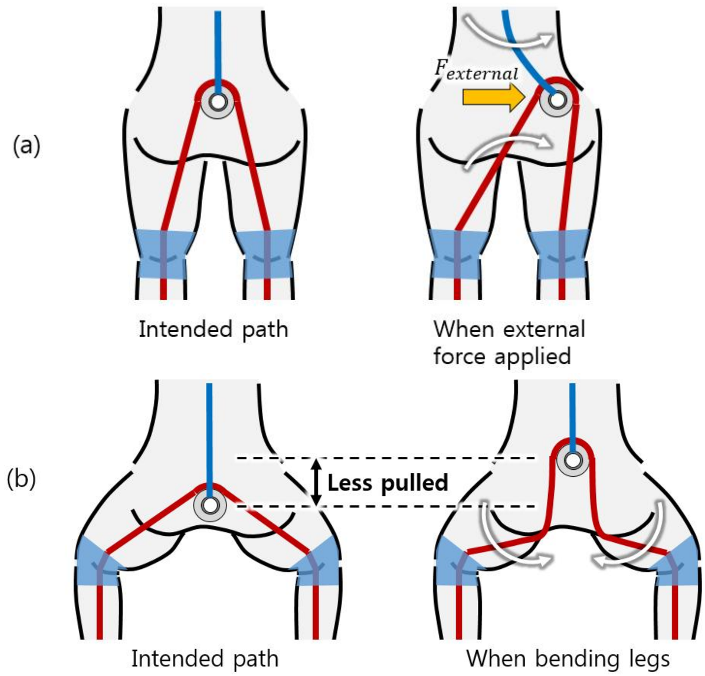


Fig. 2.15. Cable path maintaining issues. (a) When external force applied. (b) When bending legs.

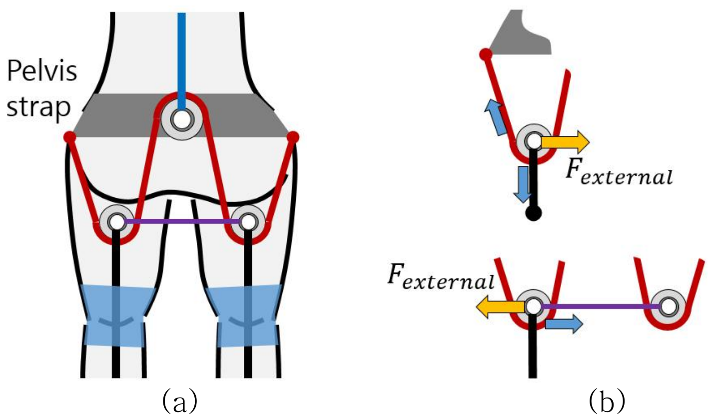


Fig. 2.16. Developed new cable path. (a) Cable path design. (b) Path maintaining principle from the horizontal external forces.

Chapter 3. Experiment

3.1. Effect of Lifting Technique

Whether the wearer is doing stoop lift or squat lift, the cable only serves to pull the shoulders and legs to back side. If so, why is the stooping motion limited whereas the squatting motion is assisted, rather than limited? To find the answer, we planned the following experiment. The subject wears the wearable device and attaches a marker at the shoulder position. It was also set to measure the tension through the load cell (333FDX-100kg, Ktoyo Co., Ltd., Korea) attached to the cable behind the back. The subject was instructed to lift a 12kg box with the handle (box size: 450*285*265mm, handle height: 220mm) in two different lifting techniques: squatting and stooping (Fig. 3.1). Each motion is recorded as a video for marker tracking (ProAnalyst, Xcitex Inc., USA). Figure 3.2 shows the result of the experiment.

First, let's compare the result of the lowering motion expressed by the solid line. When lowering the body taking a stooped motion, the tension increases sharply as the shoulder height is lowered from the initial position as shown by the red solid line. As a result, the tension became too great before reaching the target position where the handle of the box is, and the height of the shoulder could no longer be lowered. On the other hand, when the body is lowered by the squatting motion, it is observed that the tension increases more slowly as shown by the blue solid line. As a result, it was possible to reach the target position without any problem, while receiving the muscle power assistance by the tension. To sum up, this device seems to provide a different stiffness depending on the lifting technique that the wearer takes.

Next, we compare the results of the lifting motion represented by the dotted line. In the stooping motion, the tension of the cable is less at the lifting process than at the lowering process. It is a predictable result considering the hysteresis of the device. In the squatting motion, however, it is interesting to note that the lifting process produces greater tension than the lowering process. From the conclusion, this phenomenon seems to depend on the skill of the wearer of the device. The lifting motion is a movement that moves the hip joint and the knee joint simultaneously. In a squat position, if the hip joint is extended earlier than the knee, the tension in the cable will soon disappear. On the contrary, if wearer starts extending knees first, it is able to lift object while keeping the cable tight. If the wearer is familiar with the wearable device and controls the movement well, it is also possible to create a larger tension in the lifting as shown in figure 3.2. Even if wearer moves the knee joint first, the hip joint is pulled by the cable, which results in moving the knee and the hip joint at the same time.

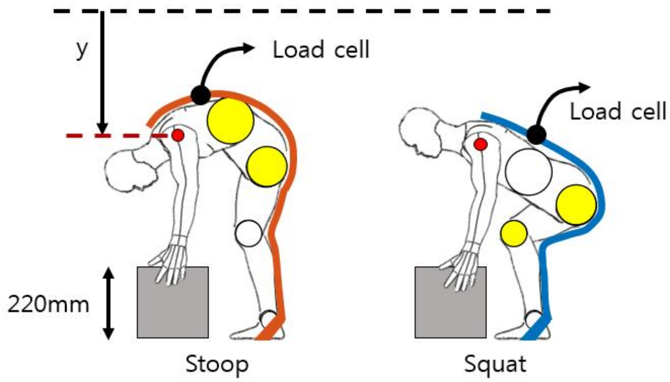


Fig. 3.1. Experiment to compare the device action according to lifting technique.

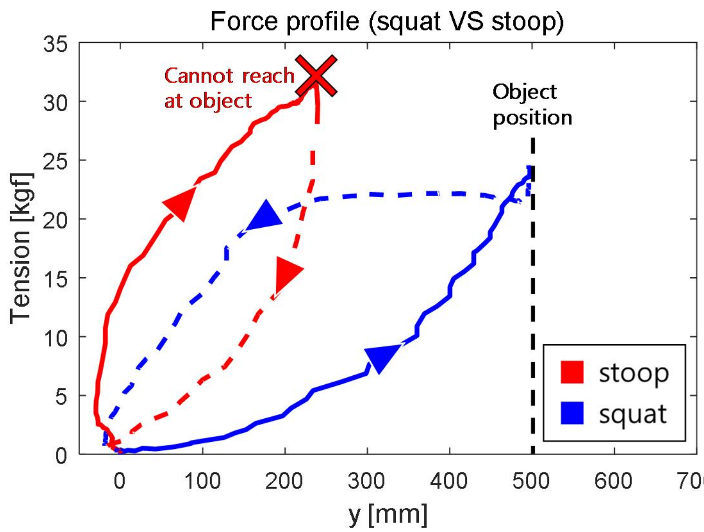


Fig. 3.2. Result of the tension when taking squat posture and stoop posture. (Solid line means lowering process and dotted line means lifting process.)

3.2. Effect of Knee Posture in the Squat

This device is designed to supply much force when the wearer pulls the knee backward while squat lift. To confirm this effect, the following experiment was performed. The subject was instructed to lift a 12kg box with the handle (box size: 450*285*265mm, handle height: 220mm) four times in a squat form. In each of the four attempts, the subject took knee posture differently. In this process, the tension was measured by attaching load cells (333FDX−100kg, Ktoyo Co., Ltd., Korea) to four different positions of cable as shown in figure 3.3.(a). In addition, markers were attached to various positions including ankle, knee, hip, and shoulder as shown in figure 3.3.(b, c). We could estimate the posture of the subjects by recording and tracking the markers (ProAnalyst, Xcitex Inc., USA). The four postures taken during the experiment were as shown in figure 3.4, with the knee in the pose #1 being the most backward, and the more protruded to forward as pose number increases. At pose #4, the knee was most protruded to forward. Figure 3.5 shows the ankle, knee, hip joint angles and the length of the knee protrusion from the tip of the foot estimated from the marker tracking results. The tension measured in each pose is shown in figure 3.6. We calculated the torque of each joints using the measured data. The torque values acting on the L_4/L_5 disc(τ_{L_4/L_5}), hip joint(τ_{hip}), and knee joint(τ_{knee}) can be obtained through the following equations:

$$\tau_{L_4/L_5} = F_A \cdot D_{L_4/L_5} \tag{4}$$

$$\tau_{hip} = 2F_B \cdot R_{hip} \tag{5}$$

$$\tau_{knee} = 2\{(F_B - F_c) \cdot R_{lever} \cdot sin(\theta_{lever/rubber}) - F_D \cdot R_{knee}\} \qquad (6)$$

Where F_A, F_B, F_C, F_D are tensions of cable at each position represented in figure 3.3.(a). $\theta_{lever/rubber}$ is angle between the lever arm and rubber string, which is calculated by marker tracking. D_{L_4/L_5} is 70mm, which is a distance between L_4/L_5 disc and cable [34].

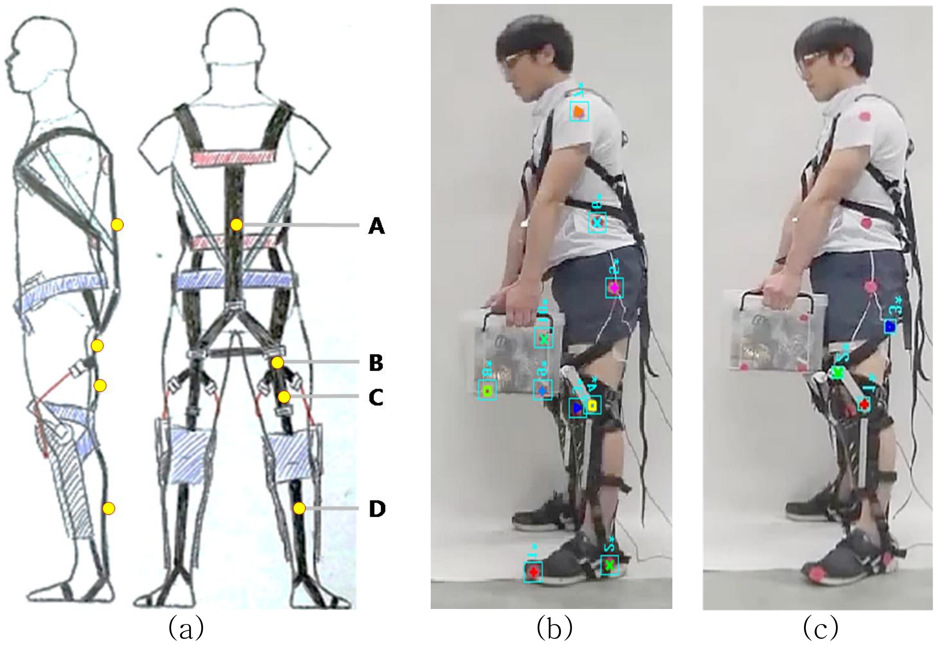


Fig. 3.3. Experiment setup for comparing assistive torque according to the knee posture. (a) Load cell position. (b) Marker positions for joint angle tracking. (c) Marker positions for rubber string angle tracking.

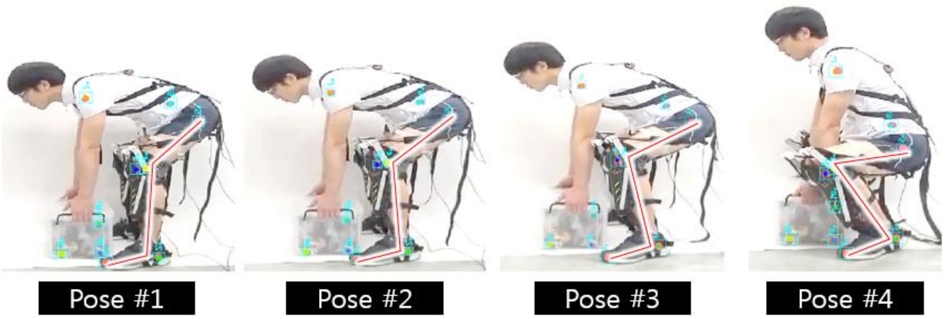


Fig. 3.4. Four postures taken during the experiment.

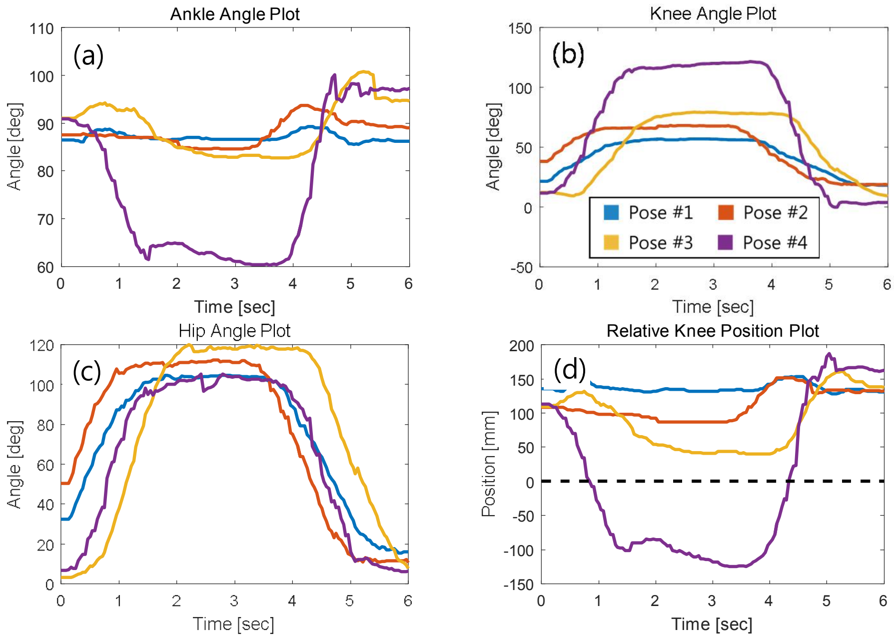


Fig. 3.5. Joint angle profiles during one lift cycle. (a) Ankle joint angle. (b) Knee joint angle. (c) Hip joint angle. (d) Relative knee horizontal position respect to the toes. (Positive value means that the knee is located behind of the toes.)

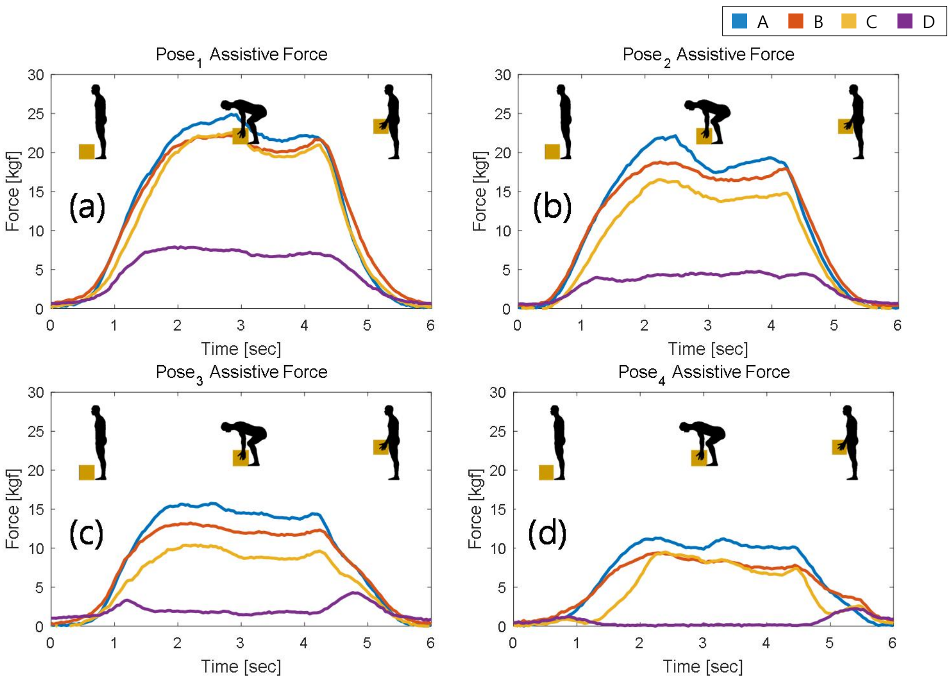


Fig. 3.6. Tension profile at the four load cell positions during one lift cycle. (a) Pose #1. (b) Pose #2. (c) Pose #3. (d) Pose #4.

Figure 3.7 shows torque generated by the wearable device for each joint during the lifting process. It can be seen that when the squat motion proceeds with the knee kept in position without protruding forward, a greater torque is generated at spine and hip joint. However, the torque generated on the knee is slightly different. This is because the tension of the inelastic cable acts as a negative torque in the position of the knee. To reduce the negative factor, we used a design that uses a rubber string to move the tension behind the knee forward. The ideal results is that the knee joint torque is positive or at least 0 for all postures. Pose #2, 3, and 4 showed the desired result, but unfortunately the pose #1 showed a negative knee torque. This seems that because in pose #1, the knee angle is so small that the rubber string does not stretch much, but the tension that occurs on the inelastic cable is the largest. However, we cannot conclude that knee fatigue at pose #1 it the largest in comparison to other postures, because originally knee muscle does not need to consume much power at the knee backed posture. It is confusing that the knee torque data is different than expected, but we still think that the wearer will be induced to take action between pose #1 and pose #2.

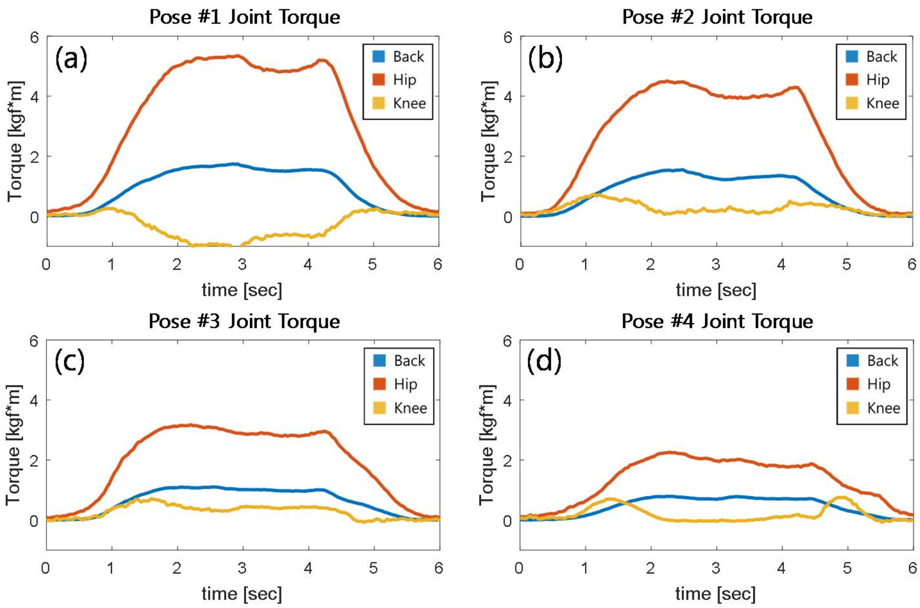


Fig. 3.7. Joint torque profiles for L_4/L_5 disc, hip joint, and the knee joint during one lift cycle. (a) Pose #1. (b) Pose #2. (c) Pose #3. (d) Pose #4.

3.3. Evaluation of Fatigue Reduction

To evaluate the effect of muscular support, the following experiment was conducted. The subject was instructed to repeat the action of lifting and lowering a 15kg box with the handle (box size: 450*285*265mm, handle height: 220mm) for 6 minutes (enough time for metabolic energy data to converge). This action was performed at a constant speed of 10 lifts/min and an interface was used to provide visual and audible signals per cycle to keep the motion speed constant (Fig. 3.8). This experiment was carried out three times in the following order: (1) squat posture without device, (2) squat posture with wearing the device, (3) stoop posture without the device. Because the subject could not bend the back while wearing the device, this process was excluded from the experiment.

After completion of each task, the subject took a rest for 20 minutes and warmed up by lifting box for 5 times. After the warm-up, additional 10 minutes of rest was taken before the next task. To measure metabolic cost and heart rate, a portable telemetry gas exchange system (K5, COSMED., Italy) was used. EMG sensors (Trigno Wireless EMG, Delsys Inc., USA) were also used to measure muscle fatigue at spinal erector, biceps femoris, and rectus femoris.

The metabolic cost and heart rate results are expressed in figure 3.9. In the case of squat lift, the peak value of metabolic energy cost decreased by 29.5% and the peak value of heart rate decreased by 13.6% when wearing our wearable device. Compared with stoop lift operation, the peak value of the metabolic energy was reduced by 24.7% and the peak value of heart rate decreased by 7.3% when the subject takes squat motion with wearing a device.

The EMG signal result is expressed in figure 3.10. The EMG data were processed in a root-mean-square with a period of 100ms. For the EMG signal, averages of the values measured in both right and left muscles were used. In the case of squat lift, the average EMG value for spinal erector, biceps femoris, and rectus femoris were decreased by 14.0%, 17.0%, and 14.3% respectively. Compared with stoop lifting operation, the average EMG value of biceps femoris was decreased by 22.3%, but value of spinal erector and rectus femoris were increased by 2.8% and 26.7% respectively. This seems to be the result due to the different kinds of muscles are commonly used for squat and stoop motion.

These results suggest that the wearable device can reduce the work fatigue by supporting enough strength support for the wearer.

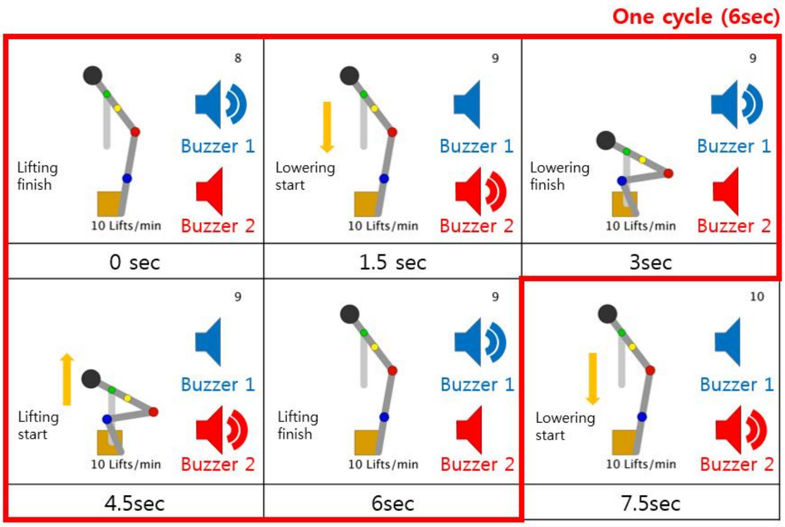


Fig. 3.8. Repetitive lifting experiment guide interface (video capture).

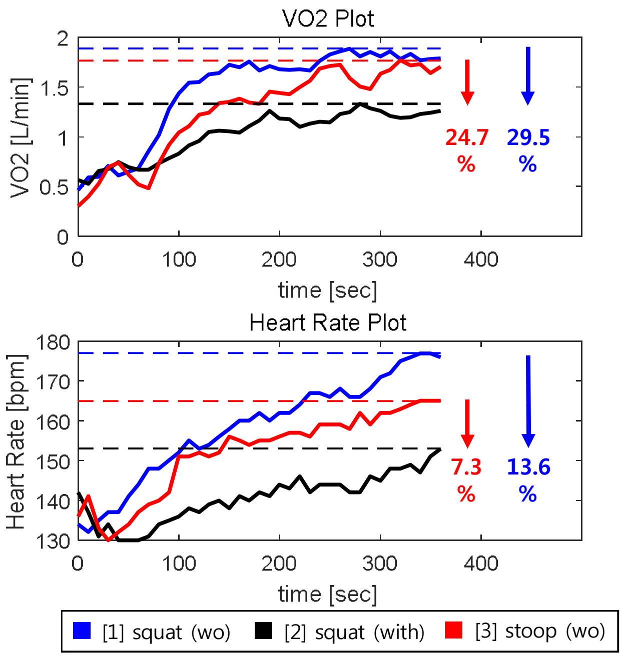


Fig. 3.9. Metabolic cost (VO2) and heart rate result.

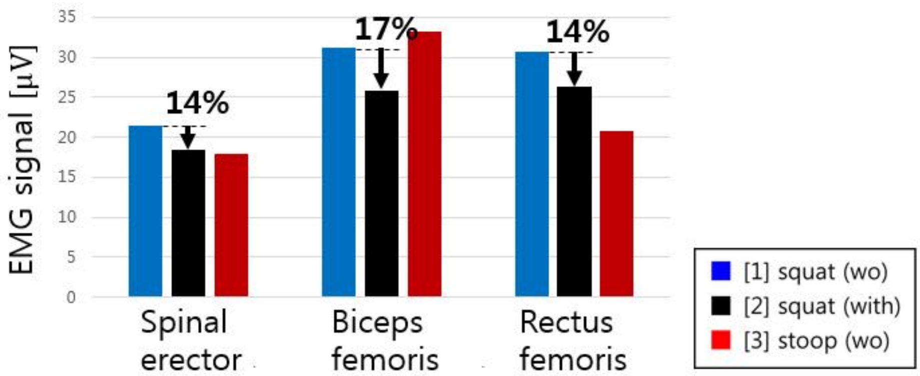


Fig. 3.10. EMG signal result.

Chapter 4. Conclusion

Generally, the human feels more comfortable with taking a bad posture than a correct posture when lifting a load. Until now, various lift assist devices have been developed but have not changed this characteristic of human. Therefore, there was a limit to changing people's lifting habits. The goal of our research is to solve the musculoskeletal disorder problems in the lifting process by inducing people to take the correct posture, which means to keep the back straight and the knee behind the toes. We have developed a wearable device that changes the characteristic of the human body system to feel comfortable at correct posture and feel uncomfortable at incorrect posture, so induce people to lift in a correct posture. Inelastic cables were used in this device as artificial ligament and muscle to constrain the human posture and support muscular force.

There were several issues in the process of developing this device. First, in order to straighten the back regardless of the wearer's intention, torque has to be created at not only lumbar spine but also all around the torso including the thoracic and shoulder. To be able to do this, an anchoring part is needed that can apply forces asymmetrically with respect to the shoulder while keep-ing the device from slipping off the body, but there was no proper example design to refer. Second, using an inelastic cable can inter-fere not only incorrect lifting motion but also other leg motions, so stuff transportation work can be difficult. Third, because the human body is not flat but curved, it was a big issue to keep the cable from getting out from the path when the cable is taut. We solved these issues by developing novel design and mechanisms. As a result of experimenting with lowering and lifting with various postures, it has been concluded that the device induces the wearer to take the cor-rect posture by giving a lot of muscular assistance when taking the correct posture and decreasing the amount of assistance as the posture differs from it. Another experiment that repeatedly lifting a heavy object over a long period of time resulted in a decrease in metabolic energy, heart rate, and EMG signal. This proved the muscular force assist effect of the device.

The device we have developed works well, but we think that design improvements will be needed for commercialization. First, the rigid materials, which are partially used for anchoring structure increase the weight of the device. And, although they do not inter-fere correct lifting motion, they cause discomfort in the everyday life activities (e.g., they hit nearby objects, and they press a body when the wearer sit on the chair). Second, it takes a long time to wear because the arrangement of the cables is complicated. In our next study, these issues will be solved and we will conduct experi-ments with numerous subjects.

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국문 초록

리프팅에서의 근력 보조와 자세 가이드를 위한 유연한 착용형 장치 개발

편한 자세는 나쁜 자세라는 말이 있다. 사람은 인체 구조로 인한 특성 상 등을 구부려서 바닥에 놓인 물체를 잡을 때 등을 펴고 다리를 구부려서 잡는 것 보다 에너지를 적게 소모하고 편하게 느낀다. 하지만 척추의 관점에서 볼 때 등을 구부리는 자세로 무거운 물체를 드는 것은 디스크 부상 위험을 2~4 배까지 증가시킬 수 있는 위험한 행위이다. 우리는 이러한 모순된 인체 시스템을 재구성함으로써 바른 자세를 편하게, 나쁜 자세를 불편하게 느끼게끔 만들어 줄 수 있는 리프트 보조용 유연한 착용형 장치를 개발했다. 본 장치의 대표적인 자세 교정 기능은 착용자가 등을 구부리는(스툽 리프트 자세) 대신 다리를 사용하는(스쿼트 리프트 자세) 리프트 테크닉을 사용하도록 리프트 습관을 바로 잡아 주는 것과, 착용자가 다리를 구부리는 과정에서 무릎을 발끝보다 뒤로 뺀 자세를 유지하도록 유도해주는 것이다. 자세를 바꿔가며 실험을 해본 결과 본 장치가 스툽 동작은 방해를 하는 반면, 스쿼트 동작은 보조해준다는 것을 알 수 있었다. 또한, 다리를 구부리는 과정에서 무릎을 뒤쪽으로 뺄수록 장치의 근력 보조 효과가 커지는 것을 확인했다. 그리고 15kg 의 상자를 10lifts/min 의 속도로 반복적으로 6 분간 들어올리는 실험을 수행하였다. 스쿼트 리프트 자세로 실험을 진행한 결과, 디바이스를 착용하면 그렇지 않을 때와 비교해서 척추 기립근, 대퇴이두근, 대퇴직근의 EMG 값을 각각 14.0%, 17.0%, 14.3%씩 감소시킬 수 있었다. 또한, 대사 에너지 소모량은 29.5%, 심장 박동수는 13.6%씩 감소시켜주는 효과가 있었다. 이를 통해 본 디바이스가 작업자들의 피로도를 감소시켜 주는 것에 효과적임을 알 수

있다. 스툽 자세로 리프트 하는 경우와 비교해 보았을 경우, 디바이스를 입고 스쿼트 자세를 취할 때 대사 에너지 소모량이 24.7%, 심장 박동수가 7.3%씩 더 작게 나오는 것을 확인했다. 따라서, 디바이스는 운송 관련 업종의 작업자들로 하여금 적은 피로로 바른 자세를 취하며 리프트 작업을 하도록 도움을 줄 수 있을 것으로 기대된다.

주요어 : 소프트 로봇, 웨어러블 로봇, 리프팅, 자세 교정, 근력 보조, 대사 에너지 소모량

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