Title of Thesis

Design of Wearable Power Assist Wear for Low Back Support Using Pneumatic Actuators

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Outline of Thesis

In this thesis, starting from the aging society’s needs of power assist robotic technology, and according the idea of designing power assist device like normal clothing, a new wearable power assist wear for low back support using pneumatic actuators is proposed. Firstly, I analyze the biomechanical model of the human spine, and get to know the main reason of LBP. Based on this analysis, two types of pneumatic actuators are selected to support the human’s back from two aspects: increasing the support force and related lever arm length. According the requirement of wearable system for measuring the human’s action, inertial sensors are selected by using sensor fusion technology. In order to provide efficient assistance force, the related appropriate control strategies is determined to minimize the interference of human actions and the assistance effectiveness of the proposed device has been proven through experiments.

This thesis is divided into seven parts.

Chapter 1 talks about aging society’s needs of power assist robotic technology, and related research in the world. Based on this demand for wearable assist robot, according the idea of designing power assist device like normal clothing, a new wearable power assist wear is proposed by using pneumatic actuators. The principles of designing a wearable power assist wear are determined: safety, user-friendliness, convenient to operate, and has a low costs.

In Chapter 2, the biomechanical model of the human spine is analyzed, and get to know the main reason of LBP: when the human is bending forward and lifting a load, for the erector spinae muscles have a small lever arm, the spine has to bear a large amount of force, which is several times of body weight. Once the disc compression force and shear force are over the limited scope that the spine can bear, permanent damage will result and lead to LBP. In order to provide the power assistance, it can be done to support the human’s back from two aspects: increasing the assistance force and related lever arm length. This is determined as the configuration of device and its operation: Using pneumatic rubber artificial muscle which is fixed parallelly to the erector spinae muscles, it can provide external assistance force, and will increase the effective amount of torque; Using layer type pneumatic actuator, it can increase the lever arm length of assistance force; At the same time, due to the expansion, it will tighten the garment and then increase the intra-abdominal pressure (IAP), which can lead to increased stability of the spine.

In Chapter 3, according to the analysis of former chapter, the key point of designing a power assist wear is to choose the appropriate actuators to output the assistance force. At first,
McKibben-type pneumatic rubber artificial muscles were chosen as actuator A to provide the assistance force, and subsequently it is improved by using elongation type pneumatic rubber artificial muscles as actuator A. For this type of muscles have a larger contraction rate, its properties are more similar to biological muscles and is better to be used for low back support. The related experiments have proved its characteristics.

In Chapter 4, I talk about the wearable Inertial Measurement Unit (IMU) sensor system. For measuring the human’s motion, inertial sensors are selected based sensor fusion technology by combine accelerometer sensors and gyro sensors. In order to verify the precision of wearable IMU sensor system for posture detection, I use Quick MAG system to detect human body's posture and compare the difference of two methods.

In Chapter 5, the effectiveness of the device have been verified through related experiments. First the appropriate control strategies were determined to minimize the interference of human actions; for the power assist wear is driven by elongation-type pneumatic rubber artificial muscles which is set parallel to the erector spinae muscles, so the sEMG activity of erector spinae muscles were measured in order to verify the effectiveness of the assistance; in order to determine the wearer's lifting capacity, related floor to waist lifting test was conducted on PrimusRs (BTE Technologies, USA) test platform; in order to provide the desired assistance force efficiently, related experiments were conducted for measuring the characteristics of human lifting action by using a force plate (ToMoCo-FPm), from the experiment the possible control method is obtained.

In Chapter 6, in order to get the assessment of the stability of human action when wearing the device, static holding test was conducted on force plate to measure the body’s movement of the center of gravity (COG), the result of experiment has shown that this device can be effective for human stability. This power assist wear can reduce the incidence of the LBP, but also can improve the wearer's stability.

Chapter 7 is summary of the whole thesis, and in order to improve its performance in the future, further improvements are discussed.
Design of Wearable Power Assist Wear for Low Back Support
Using Pneumatic Actuators

Abstract

This research focuses on developing a safe, lightweight, power assist device that can be worn by people during lifting or static holding tasks to prevent them from experiencing low back pain (LBP). In consideration of their flexibility, light weight, and large force to weight ratio, two types of pneumatic actuators are employed in assisting low back movement for their safety and comfort. At first, McKibben-type pneumatic rubber artificial muscles were chosen as actuator A to provide the assistance force, and subsequently it is improved by using elongation type pneumatic rubber artificial muscles as actuator A. Actuator A is installed in the outer layer of the garment, and its two ends are fixed on the shoulders and thighs. It can output contractile force, assisting the erector spinae muscles in the same direction. Compared to McKibben-type pneumatic rubber artificial muscle, the elongation type has a larger contraction rate. Actuator B is a layer-type of pneumatic actuator; it is composed of two balloons, and it is installed in the inner layer of the garment. The biomechanical model of the human spine is analyzed, and get to know the main reason of LBP: when the human is bending forward and lifting a load, for the erector spinae muscles have a small lever arm, the spine has to bear a large amount of force, which is several times of body weight. By taking into account the biomechanic structure of the human spine, this device can provide support in two ways. Actuator A acts as an external muscle power generators to reduce the force requirement for the erector spinae muscles. As actuator B acts as a moment arm of the contractile force generated by actuator A, it will increase the effective amount of torque. The device can be worn directly on the body like normal clothing. Because there is no rigid exoskeleton frame structure, it is lightweight and user friendly. The system's Inertial Measurement Unit (IMU), composed of accelerometer sensors and gyro sensors to measure the human motion signals, can monitor the angles of the human body in real-time mode. By measuring the EMG signal of the human erector spinae muscles, the assistance effectiveness of the proposed device has been proven through experiments.
## CONTENTS

Outline of Thesis..................................................................................................................................... I
Abstract.................................................................................................................................................. I
List of Figures.......................................................................................................................................... III

1 Introduction........................................................................................................................................ 1
   1.1 Aging society.................................................................................................................................. 1
   1.2 Power assist robot......................................................................................................................... 1
   1.3 Power assist device for low back support..................................................................................... 3
   1.4 Power assist wear for low back support using pneumatic actuator............................................... 5
   1.5 Main content of thesis.................................................................................................................... 6

2 Design principles of power assist wear for low back support............................................................... 9
   2.1 Low back pain............................................................................................................................... 9
   2.2 Low back biomechanics for lifting................................................................................................ 9
   2.3 Configuration of device and its operation..................................................................................... 11
   2.4 Conclusion.................................................................................................................................... 14

3 Actuators of system.............................................................................................................................. 15
   3.1 McKibben-type pneumatic rubber artificial muscle...................................................................... 15
   3.2 Elongation-type pneumatic rubber artificial muscle..................................................................... 16
   3.3 Layer-type pneumatic actuator.................................................................................................... 19
   3.4 Conclusion.................................................................................................................................... 20

4 Wearable IMU sensor system................................................................................................................. 21
   4.1 Human motion detection............................................................................................................... 21
   4.2 Wearable IMU sensor system......................................................................................................... 21
   4.3 Experiment in Quick MAG system............................................................................................... 24
   4.4 Conclusion.................................................................................................................................... 26

5 Evaluation of assistance effectiveness.................................................................................................. 27
   5.1 Device control strategy.................................................................................................................. 27
   5.2 Evaluation of assistance effectiveness using sEMG signals......................................................... 29
   5.3 Floor to waist lifting test on PrimusRs......................................................................................... 32
   5.4 Improvement on control strategy.................................................................................................. 33
   5.5 Conclusion.................................................................................................................................... 36

6 Static holding test................................................................................................................................ 37
   6.1 Static holding test.......................................................................................................................... 37
   6.2 Test result...................................................................................................................................... 38
   6.3 Conclusion.................................................................................................................................... 40

7 Conclusion and future developments.................................................................................................... 41
List of Figures

Fig. 1 HAL (Hybrid Assistive Limbs) series: HAL-1, HAL-3, HAL5[2] ........................................ 2
Fig. 2 Power assist suit[4] ........................................................................................................... 2
Fig. 3 Wearable muscle suits[6] ................................................................................................. 2
Fig. 4 A wearable exoskeleton power assist device using DC motor[9] ...................................... 3
Fig. 5 On-body lift assistive device (PLAD)[11] ......................................................................... 3
Fig. 6 Smart suit for horse trainers[12] .................................................................................... 4
Fig. 7 Smart Suit Lite[13] ........................................................................................................... 4
Fig. 8 Power assist wear for low back support .......................................................................... 6
Fig. 9 Cantilever model of spine in lifting ............................................................................... 9
Fig. 10 Wearable power assist device using curved pneumatic rubber artificial muscles. 11
Fig. 11 Mechanism of power assist wear for low back support .............................................. 12
Fig. 12 Simulation of reduced compression force on L5/S1 disc ............................................... 13
Fig. 13 Power assist wear for low back support (2011) .......................................................... 15
Fig. 14 Elongation-type pneumatic rubber artificial muscle .................................................... 16
Fig. 15 Relation between supplied pressure and displacement .............................................. 17
Fig. 16 Relation between displacement and contractile force .............................................. 17
Fig. 17 Overview of TPU balloon .......................................................................................... 19
Fig. 18 Relation between expansion force and displacement in height .................................. 19
Fig. 19 Structure of layer-type pneumatic actuator ................................................................. 20
Fig. 20 Sensor fusion algorithm ............................................................................................... 22
Fig. 21 IMU sensor composed of gyro sensors and accelerometer units .................................. 23
Fig. 22 Orientation of axes of sensitivity and polarity of rotation ........................................ 23
Fig. 23 Definition in terms of human joints .......................................................................... 23
Fig. 24 Experiment in Quick MAG system ............................................................................. 24
Fig. 25 Five-segment model .................................................................................................... 25
Fig. 26 $\theta_H$, $\theta_K$, $\theta_L$ angles of human body during the lifting cycle .............................. 25
Fig. 27 Layer-type pneumatic actuator pressure response during flexion motion .................. 27
Fig. 28 Structure of control system ...................................................................................... 28
Fig. 29 Block diagram of control system .............................................................................. 28
Fig. 30 Lifting load experiment .............................................................................................. 29
Fig. 31 Evaluation by measuring sEMG signal of erector spinae muscles (loads: 0 kg).......... 30
Fig. 32 Evaluation by measuring sEMG signal of erector spinae muscles (loads: 12.6 kg)...... 31
Fig. 33 Floor to waist lifting test on PrimusRS ...................................................................... 32
Fig. 34 Maximum lifting weight and output power ............................................................... 32
<table>
<thead>
<tr>
<th>Fig.</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>35</td>
<td>Pitch angles of body during lifting cycle</td>
</tr>
<tr>
<td>36</td>
<td>The angle to pressure algorithm in one cycle</td>
</tr>
<tr>
<td>37</td>
<td>Spine stability requires agonist-antagonist co-activation</td>
</tr>
<tr>
<td>38</td>
<td>Static holding test on force plate</td>
</tr>
<tr>
<td>39</td>
<td>The moving length and velocity distribution of the COG</td>
</tr>
</tbody>
</table>
1 Introduction

1.1 Aging society

With the higher average life expectancy and the declining birthrate, the world's developed countries, including Japan, have entered the aging society. According to the statistic data of the Ministry of Internal Affairs and Communications of Japan, the elderly share (age 65 and above) of the population is 24.1% now, and will reach 3473 million population in 2025, account for 28.7% of the total population.

With the rapid arrival of the aging society, the demand for professional caregivers has increased drastically. The professional caregiver shortage has become a profound social problem. It has become a major focus in recent years; researchers have been using robotic technologies to develop many kinds of assistive and rehabilitative devices for people with disabilities or to develop medical devices used by caregivers. Wearable robot is a such kind of device that can be defined as those worn by human operators, whether to supplement the function of a limb or to replace it completely[1].

In order to prepare for such upcoming problems as labor shortage of caregivers and create a new industries on nursing care robot, the Ministry of Health, Labor and Welfare and the Ministry of Economy of Japan, have published four priority areas in the use of nursing care robot in 2012. One of the priority areas are the development of wearable power assist devices and non wearable lifting assist devices used in nursing care fields.

1.2 Power assist robot

From the early 1990s, related research of wearable assist devices which focused on health care or support in daily life have begun to be done in some universities of Japan. The assistance effectiveness of most wearable robots now under research or already developed have been verified by related experiments. Prof. Sankai developed the Hybrid Assistive Limbs (HAL) in order to enhance and upgrade the human capabilities (Fig. 1)[2][3]. Prof. Yamamoto developed a wearable power assist suit using pneumatic actuators aimed at giving caregivers the extra strength to lift patients while avoiding back injuries (Fig. 2)[4]. Prof. Kobayashi developed a muscle suit using McKibben-type rubber artificial muscles to help users normally needing assistance to move unaided (Fig. 3)[5][6]. These wearable robots have multiple degrees of freedom and can provide assistance for whole body movement, but they are not suitable for use in small spaces due to their rigid frame and considerable weight.
Fig. 1  HAL (Hybrid Assistive Limbs) series: HAL-1, HAL-3, HAL5[2]

Fig. 2  Power assist suit[4]

Fig. 3  Wearable muscle suits[6]
1.3 Power assist device for low back support

In Japan, due to the flood of retirements of Japan’s baby boom generation (dankai no setai), from 2012 to 2014 the annual retired old people will be more than one million. The corresponding term of caregivers will increase from 1.33 millions to more than 1.5 times (2.32-2.44 millions) to meet the requirements. Most of caregivers have high prevalence rates of low back pain (LBP) and a high incidence of worker’s compensation claims for back injuries[7].

Fig. 4  A wearable exoskeleton power assist device using DC motor[9]

Fig. 5  On-body lift assistive device (PLAD)[11]
For the purpose of power assistance for low back support, some research groups are developing human motion assistance devices that just focus on the waist area. An wearable exoskeleton power assist system for low back support was built by using DC motor (Fig. 4) [8][9]. On-body lift assistive device (PLAD) was developed to reduce force requirements of back muscles by using six elastic elements anchored at shoulders and knees (Fig. 5)[10][11]. Smart Suit was developed to provide the assist force using the elastic materials and controlled by adjusting the length of the elastic materials by using DC motor (Fig. 6)[12]. And a passive power assist device, Smart Suit Lite was developed in 2011, which is a compact, lightweight power assist device that utilizes the elastomeric force of elastic materials (Fig. 7)[13]. These
devices can be divided into two types: active power assist devices and passive power assist devices. As for the actuator using motor, its flexibility is not good comparing with pneumatic actuator. As for the system above mentioned using elastic element as actuator, the elastic element store energy of human during the upper body is lowered and then feedback the stored energy during the upward phase, it is passive assist type or semi-active type. Most of them cannot adjust the assist power during different task and the output assistance is insufficient.

1.4 Power assist wear for low back support using pneumatic actuator

For designing a wearable power assist device, the safety and user-friendliness are important factors that should be considered.

In Prof. Noritsugu’s laboratory (MCRLAB, Mechanical Control and Robotics Laboratory), a variety of power assist devices are developed by using pneumatic actuators for its advantages of softness, high power-to-weight ratio, and low cost[14]-[17]. The idea of power assist wear is initially proposed by Prof. Noritsugu in the world. Using pneumatic rubber artificial muscles, these devices can be fitted to the human body like normal clothing to assist the power of muscles that support the activities of daily living, rehabilitation, training[18].

Based on previous researches, in this study a wearable power assist wear for low back support is proposed using pneumatic actuators, by taking the requirement for light weight and human body wearing comfort into consideration. This device can transfer assistance force to the shoulders and thighs without using an exoskeleton structure. The total weight of the prototype (Fig. 1) is 0.8 kg. The garment is made based on a lumbar support belt and is equipped with two types of pneumatic actuators. Actuator A, installed on the outer layer of the garment, has its ends fixed at the shoulders and thighs. Actuator A consists of five elongation-type pneumatic rubber artificial muscles arranged in parallely. It can output contractile force, assisting erector spinae muscles in the same direction of motion. Actuator B is installed in the inner layer of the garment. It is a layer-type pneumatic actuator composed of two balloons of thermoplastic polyurethane (TPU) materials.

Comparing to the power assist device of using DC motor as actuator, this device is safe, light weight, easy to wear and operate, without the hidden dangers of electric shock. Comparing to the passive power assist device of using elastic materials, this device can output a large assistance force, and the output assistance can be adjusted by supplying the pneumatic actuators with different pressure.
1.5 Main content of thesis

In this thesis, starting from the aging society’s needs of power assist robotic technology, and according the idea of designing power assist device like normal clothing, a new wearable power assist wear for low back support using pneumatic actuators is proposed. Firstly, I analyze the biomechanical model of the human spine, and get to know the main reason of LBP. Based on this analyze, two types of pneumatic actuators are selected to support the human’s back from two aspects: increasing the support force and related lever arm length. According the requirement of wearable system for measuring the human’s action, inertial sensors are selected by using sensor fusion technology. In order to provide efficient assistance force, the related appropriate control strategies is determined to minimize the interference of human actions and the assistance effectiveness of the proposed device has been proven through experiments.

This thesis is divided into seven parts.

Chapter 1 talks about aging society’s needs of power assist robotic technology, and related research in the world. Based on this demand for wearable assist robot, according the idea of designing power assist device like normal clothing, a new wearable power assist wear is proposed by using pneumatic actuators. The principles of designing a wearable power assist wear are determined: safety, user-friendlyness, convenient to operate, and has a low costs.

In Chapter 2, the biomechanical model of the human spine is analyzed, and get to know the main reason of LBP: when the human is bending forward and lifting a load, for the erector spinae muscles have a small lever arm, the spine has to bear a large amount of force, which is
several times of body weight. Once the disc compression force and shear force are over the limited scope that the spine can bear, permanent damage will result and lead to LBP. In order to provide the power assistance, it can be done to support the human’s back from two aspects: increasing the assistance force and related lever arm length. This is determined as the configuration of device and its operation: Using pneumatic rubber artificial muscle which is fixed parallelly to the erector spinae muscles, it can provide external assistance force, and will increase the effective amount of torque; Using layer type pneumatic actuator, it can increase the lever arm length of assistance force; At the same time, due to the expansion, it will tighten the garment and then increase the intra-abdominal pressure (IAP), which can lead to increased stability of the spine.

In Chapter 3, according to the analysis of former chapter, the key point of designing a power assist wear is to choose the appropriate actuators to output the assistance force. At first, McKibben-type pneumatic rubber artificial muscles were chosen as actuator A to provide the assistance force, and subsequently it is improved by using elongation type pneumatic rubber artificial muscles as actuator A. For this type of muscles have a larger contraction rate, its properties are more similar to biological muscles and is better to be used for low back support. The related experiments have proved its characteristics.

In Chapter 4, I talk about the wearable Inertial Measurement Unit (IMU) sensor system. For measuring the human’s motion, inertial sensors are selected based sensor fusion technology by combine accelerometer sensors and gyro sensors. In order to verify the precision of wearable IMU sensor system for posture detection, I use Quick MAG system to detect human body's posture and compare the difference of two methods.

In Chapter 5, the effectiveness of the device have been verified through related experiments. First the appropriate control strategies were determined to minimize the interference of human actions; for the power assist wear is driven by elongation-type pneumatic rubber artificial muscles which is set parallel to the erector spinae muscles, so the sEMG activity of erector spinae muscles were measured in order to verify the effectiveness of the assistance; in order to determine the wearer’s lifting capacity, related floor to waist lifting test was conducted on PrimusRs (BTE Technologies, USA) test platform; in order to provide the desired assistance force efficiently, related experiments were conducted for measuring the characteristics of human lifting action by using a force plate (ToMoCo-FPm), from the experiment the possible control method is obtained.

In Chapter 6, in order to get the assessment of the stability of human action when wearing the device, static holding test was conducted on force plate to measure the body’s movement of the center of gravity (COG), the result of experiment has shown that this device can be effective for human stability. This power assist wear can reduce the incidence of the LBP, but
also can improve the wearer's stability.

Chapter 7 is summary of the whole thesis, and in order to improve its performance in the future, further improvements are discussed.
2 Design principles of power assist wear for low back support

Due to most of health caregivers troubled by low back pain problems, the requirement for low back support is needed. So it is critical to find the cause of LBP when designing power assist wear for low back support.

2.1 Low back pain

LBP is common in various occupations, its presence being related to activities requiring repetitive lifting, repeated activities in bending forward positions, and with a high energetic load[19]. Related results also suggest that bending activities involving higher degrees of trunk flexion were associated with disabling types of LBP in certain working populations[20]-[23]. Such work characteristics are common among nursing caregivers. The prevalence of LBP in nursing is high in comparison with other occupations and in relation to other types of work. Risk factors include physical work such as manual lifting and transferring of patients, working conditions such as working time and rest during the night shift, and the working environment. Among these factors, exposures to frequent manual lifting and transferring of patients were widely recognized factors.

2.2 Low back biomechanics for lifting

![Cantilever model of spine in lifting](image)

Fig. 9 Cantilever model of spine in lifting
As the low back is its most vulnerable to injury during manual lifting activities and transferring, lifting has most frequently been associated with LBP. As shown in Fig. 9, when a person bends forward to lift heavy loads, the spine can be viewed as a cantilever pivoting on the hip joint[24]. Both the weight of the upper body \( m_{\text{HAT}}g \) (including the head, arms, and trunk) and the loads \( m_Lg \) form the resistance, balanced by efforts created by Intra-abdominal Pressure (IAP), co-contraction of the trunk muscles, and external assistance[25]. All of the loads will cause significant stress on the body structures, especially at the disc between the fifth lumbar and the first sacral vertebrae (L5/S1 disc). Here the IAP is neglected because it is relatively small, usually predicted to be 10% of the total at most. According to the static equilibrium, the moment on the L5/S1 disc of the human spine can be expressed as Eq. (1).

\[
M_{\text{L5/S1}} = m_{\text{HAT}}gB + m_LgL
\]  

where \( B \) is the moment arm from the L5/S1 disc to the center of gravity (COG) of the upper body, and \( L \) is the moment arm of loads.

The force of co-contraction of erector spinae muscles \( F_M \) is simplified and assumed to be parallel to the back, so \( F_M \) can be expressed as Eq. (2).

\[
F_M = M_{\text{L5/S1}} / E
\]

where \( E \) is the related moment arm of the erector spinae muscles, and it is often given a constant value of 0.05 m.

With the inclined angle of the L5/S1 disc assumed to be \( \alpha \), the compression force \( F_C \) and the shear force \( F_S \) of the disc can be expressed by Eq. (3) and Eq. (4):

\[
F_C = (m_{\text{HAT}} + m_L)g \cos \alpha + F_M
\]

\[
F_S = (m_{\text{HAT}} + m_L)g \sin \alpha
\]

The angle \( \alpha \) is estimated to depend on the posture, as follows:

\[
\alpha = -17.5 - 0.120\theta_T + 0.230\theta_K + 0.00120\theta_T\theta_K + 0.0050\theta_T^2 - 0.00075\theta_K^2 + 40
\]

where \( \theta_T \) is the angle of the trunk, and \( \theta_K \) is the angle of the knee. All angles should be represented in degrees. From the point of view of mechanics, the spine is not ideal for lifting loads, especially in a flexion posture. Because the back muscle has a small lever arm, the spine has to bear a large amount of force, several times of body weight, and once the disc compression force and shear force are over the limited scope that the spine can bear, permanent damage will result and lead to LBP. According to the National Institute for Occupational Safety and Health (NIOSH), the maximum acceptable limit for compression force on the L5/S1 disc is 3400 N[26]. Related research has shown that introducing additional assistance for the trunk muscles can enhance the stability of the spine and reduce the risk of LBP[27].
2.3 **Configuration of device and its operation**

![Image of device](image)

(a) Inner pressure: 0 kPa  (b) Inner pressure: 500kPa

Fig. 10 Wearable power assist device using curved pneumatic rubber artificial muscles

A wearable waist power assist device has been proposed in our laboratory[28]. As is shown in Fig. 10, the device, composed of curved pneumatic rubber artificial muscles and a garment, can support waist movement by supplying the curved pneumatic rubber artificial muscles with the desired pressure. The effectiveness of the assistance has been proven by related experimental data. However, the device, using a garment with metal rods as a link, will limit the range of back motion (twisting motion), and the combined weight of the device (4.5 kg) causes discomfort for the wearer. Additionally, the device uses a tilt-angle sensor to measure the movement of the torso, which is not sufficient to identify the complete body posture.

Based on previous designs, a new design is proposed, taking the requirement for lightweight and human body wearing comfort into consideration. This does not just mean providing assistance force to the hip joint, but considering the biomechanical characteristics of the human spine, assistance force parallel to the erector spinae muscles is provided. The device can transfer assistance force to the shoulders and thighs without using an exoskeleton structure. The total weight of the prototype (Fig. 8) is 0.8 kg. The garment is made based on a lumbar support belt and is equipped with two types of pneumatic actuators. Actuator A, installed on the outer layer of the garment, has its ends fixed at the shoulders and thighs. Actuator A consists of five elongation-type pneumatic rubber artificial muscles arranged in parallelly. It can output contractile force, assisting erector spinae muscles in the same direction of motion. Actuator B is installed in the inner layer of the garment. It is a layer-type pneumatic actuator composed of two balloons of thermoplastic polyurethane (TPU) materials.
Fig. 11  Mechanism of power assist wear for low back support

Fig. 11 shows the mechanism of the proposed power assist wear for low back support when the user puts it on and lifts up load. The total weight of the device is neglected in the following analysis for it is lightweight, and all the definition are assumed as same as former section described. The elongation-type pneumatic rubber artificial muscles output the assistance force $F_A$, the direction of which is simplified and assumed to be the same as the co-contraction force of erector spinae muscles $F'_M$. The moment arm of $F_A$ is defined as $A$, which can be adjusted by changing the supplied pressure of layer-type actuator B. With the assistance of the device, the reduced moment on the L5/S1 disc becomes $M'_{L5/S1}$.

$$M'_{L5/S1} = M_{L5/S1} - F_A A \tag{6}$$

The reduced co-contraction force $F'_M$ of erector spinae muscles can therefore be expressed as follows:

$$F'_M = \frac{M'_{L5/S1}}{E} \tag{7}$$

The related compression force and shear force can be expressed as in Eq. (8) and Eq. (9).

$$F'_C = (m_{HAT} + m_L)g \cos \alpha + F'_M + F_A \tag{8}$$

$$F'_S = (m_{HAT} + m_L)g \sin \alpha \tag{9}$$

Considering the above Equations, the reduced compression force can be deduced as follows:

$$F'_C = F_C - \frac{A - E}{E} F_A \tag{10}$$

The moment arm $A$ is always larger than the moment arm $E$, so it can be concluded from
the Eq. (6) and Eq. (10) that the device can reduce the required moment and compression force on the L5/S1 disc.

![Diagram of force distribution](image1.png)

Fig. 12 Simulation of reduced compression force on L5/S1 disc

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<thead>
<tr>
<th>Tab. 1</th>
<th>Assumption of simulation</th>
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<tr>
<td>Unit:</td>
<td>Value:</td>
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<tr>
<td>( m_{\text{sh}} ) (kg)</td>
<td>0.475* ( W )</td>
</tr>
<tr>
<td>( B ) (m)</td>
<td>0.17* ( H )*( \sin(\theta) )</td>
</tr>
<tr>
<td>( F_C ) (N)</td>
<td>250</td>
</tr>
<tr>
<td>( A ) (m)</td>
<td>0.13</td>
</tr>
<tr>
<td>( \theta_k ) (degree)</td>
<td>180</td>
</tr>
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For example, when a normal adult man (weight: \( W=66 \) kg, height: \( H=1.65 \) m) lifts a load (weight: \( m_2=20 \) kg) in stoop mode from the ground, which is shown in Fig. 12(a). Here assumes that the length of \( L \) is 0.4 m, the related data of human body are shown in Tab.1. During the process that the upper body is bending from 0 degree to 65 degrees, related compression force on L5/S1 disc are shown in Fig. 12(b), \( F_C \) is the compression force under the condition of without assistance force, its value increases gradually from 1955 N to 3326 N when the angle of trunk \( \theta_l \) increases from 0 degree to 65 degrees, it almost reaches to the limit of the human body can bear (3400 N). \( F'_C \) is the reduced compression force under the condition of with assistance force, its value increases gradually from 1555 N to 2926 N, which is smaller than corresponding value of \( F_C \). So in theoretically this device can provide the assistance force which can reduce the risk of LBP.
The main feature of the device is that it can provide assistance from the two aspects. Assistance force $F_A$ of actuator A acts as an external muscle power generator to reduce the output force $F_M$ provided by the erector spinae muscles, and actuator B acts as moment arm $A$ of external assistance force $F_A$, and this will increase the effective amount of torque. At the same time, due to the expansion of actuator B, which will tighten the garment and then increase the intra-abdominal pressure (IAP), it can lead to increased stability of the spine.

2.4 Conclusion

In this chapter, I analyze the biomechanical model of the human spine, and get to know the main reason of LBP: when the human is bending forward and lifting a load, for the erector spinae muscles have a small lever arm, the spine has to bear a large amount of force, which is several times of body weight. Once the disc compression force and shear force are over the limited scope that the spine can bear, permanent damage will result and lead to LBP. In order to provide the power assistance, I can support the human’s back from two aspects: increasing the assistance force and related lever arm length. This is determined as the configuration of device and its operation: Using pneumatic rubber artificial muscle which is fixed parallely to the erector spinae muscles, it can provide external assistance force, and will increase the effective amount of torque; Using layer type pneumatic actuator, it can increase the lever arm length of assistance force; At the same time, due to the expansion, it will tighten the garment and then increase the intra-abdominal pressure (IAP), which can lead to increased stability of the spine.
3 Actuators of system

According to the analyze of former chapter, the key point of designing a power assist wear is to choose the appropriate actuators to output the assistance force. Pneumatic rubber artificial muscles, which has the advantages of softness, light weight, and low cost, are particularly suitable for the field of power assist robots.

3.1 McKibben-type pneumatic rubber artificial muscle

As one of the pneumatic actuator variety, McKibben-type pneumatic rubber artificial muscle achieved the most attention and the best results. The most attractive attribute of such actuators is their high force to weight ratio. A ratio of 400:1 for McKibben-type made them a promising approach compared to alternatives at the time such as pneumatic cylinders and DC motors, which had ratios in the order of 16:1.

At first, McKibben-type pneumatic rubber artificial muscle is chosen as actuator A to develop a prototype power assist wear in 2011, which is shown in Fig. 13. Actuator A is made by five pieces of McKibben-type artificial rubber muscles which are paralleled to compose a plane actuator to provide the assistance force. This device can reduce the contraction force of back muscles by changing the pressure of the pneumatic artificial rubber muscle; it also can change the direction of supporting force by changing the pressure of layer type pneumatic actuator.

![Fig. 13  Power assist wear for low back support (2011)](image)

But the contraction rate of McKibben-type artificial rubber muscle (about 25%) is smaller than that of human muscle. In order to assist human movement efficiently, an extra mechanical structure is needed to install McKibben-type pneumatic rubber artificial muscle
and transform the output force into the desired torque[29]. These devices are usually designed as exoskeletons, which decreases their performance due to their weight, larger size, and low degree of freedom. In this device, I have to increase the pressure of Actuator B to achieve its purpose, but the contact part of human body had to bear a greater pressure force, which will decrease the body's comfort. A new type of actuator like human muscle’s characteristics should be developed for designing a power assist wear.

3.2 Elongation-type pneumatic rubber artificial muscle

Developing power assist wear that is like normal clothing requires the development of a new type of pneumatic rubber artificial muscle to meet the requirements of safety and user-friendliness. In this part, an elongation-type of pneumatic rubber artificial muscle is employed for the special requirement in assisting low back movement.

![Fig. 14 Elongation-type pneumatic rubber artificial muscle](image)

Fig. 14 shows the structure of the elongation-type pneumatic rubber artificial muscle. It is composed of a rubber tube covered with a bellows sleeve and closed at both ends with ties. The outer diameter of the rubber tube is 12.0 mm; the inner diameter is 10.0 mm. The bellows sleeve is woven of twisted fiber cord, which has a maximum expansion diameter of 14.0 mm and a minimum contraction diameter of 4.0 mm.

When first pressurized, the elongation-type pneumatic rubber artificial muscle expands slightly in the radial direction, putting the internal rubber tube into close contact with the external sleeve. As the pressure continues to rise, due to the fact that radial expansion is limited by the external sleeve, the artificial muscle expands in the axial direction, and it becomes longer. The higher pressure, the longer it becomes. Fig. 15 shows the relation between supplied pressure and $L_{d}$, related displacement of elongation-type pneumatic rubber artificial muscle, if there is no external load connected. The maximum length is attained when it is pressurized; the minimum length is attained when it is not. For this type of muscle
(original length \( L_0 = 320 \text{ mm} \)), the total length can reach 490 mm \((L_0 = 170 \text{ mm})\) when pressurized to 500 kPa.

**Fig. 15** Relation between supplied pressure and displacement

![Graph showing the relation between supplied pressure and displacement](image)

**Fig. 16** Relation between displacement and contractile force

![Graph showing the relation between displacement and contractile force](image)

Using an FGS-250PVH Motorized Test Stand (NIDEC-SHIMPO Corporation), related experiments were done to measure the corresponding relationship between length and contractile force under different constant pressures, as shown in Fig. 16. The elongation-type pneumatic rubber artificial muscle can be seen as a spring with a variable stiffness coefficient. The output contractile force can be expressed as Eq. (11).

\[
F = k(P, L_d)L_d 
\]

(11)

where \( F \) is the output contractile force, \( k \) is the stiffness coefficient of the spring model, \( P \) is the supplied pressure, and \( L_d \) is the displacement of the elongation-type pneumatic rubber artificial muscle. When it is pressurized, due to the interaction between the expanding force of the internal pressure and the elasticity of the rubber tube, the stiffness coefficient becomes larger as the pressure increases. The longer the artificial muscle, the more elastic potential energy, and the larger the output contractile force.
In Fig. 16, the output contractile force and the displacement have a linear relation in their working range \( L_{d\text{max}} = 390 \text{ mm} \). Within the scope of the working range, the output contractile force decreases as the pressure increases. The contraction rate \( \varepsilon \) of the elongation-type pneumatic rubber artificial muscle can be defined as the following equation:

\[
\varepsilon = \frac{L_d}{L_o + L_d} \quad (12)
\]

Using Eq. (12), the maximum contraction rate of the elongation-type pneumatic rubber artificial muscle can reach \( \varepsilon_{\text{max}} = L_{d\text{max}} / (L_o + L_{d\text{max}}) = 54.9\% \), which is larger than common McKibben-type pneumatic rubber artificial muscle; its properties are more similar to biological muscles[30]. The biological muscle can be stretched beyond its original length, and can also be contracted to a shorter length than its original one \( (0.7L_o < L < 1.4L_o) \). However the contraction rate of McKibben-type pneumatic rubber artificial muscles is usually only about 30% with a pressure of 500 kPa, and they cannot be stretched beyond their original length[31].

In order to assist low back movement using McKibben-type pneumatic rubber artificial muscle, a longer length will be required to obtain the contractile force needed. The artificial muscle will be longer than human back muscle, so it will be difficult to install it just along the surface of a human back. An extra mechanical structure will be needed to install and transform the output force into the desired torque. Compared to McKibben-type pneumatic rubber artificial muscle, the elongation type has a larger contraction rate, and its working principle is contrary to that of McKibben-type pneumatic rubber artificial muscles in developing power assist wear for low back support.

In this device, the elongation-type pneumatic rubber artificial muscles are used as external muscle power generators. The assistance force \( F_d \) can reduce the requirement for erector spinae muscles. When a person bends forward, the erector spinae muscles become longer to balance the weight of the person's body. As shown in Fig. 16, from state A to state B for one elongation-type pneumatic rubber artificial muscle, the output contractile force will become from 0 N to 25 N larger when it is stretched by being supplied with constant pressure of 500 kPa. The entire assistance force (consisting of five parallel muscles) \( F_d \) can reach 125 N. Here, the force is moderate; it is just to be used to balance the weight of the trunk. During the process of lifting a load, considerable assistance can be achieved by reducing the supply pressure to 0 kPa from state B to state C. As one muscle can output a contractile force reaching 50 N, the entire assistance force can theoretically reach 250 N.

Therefore, by supplying different pressures, this device can provide different amounts of assistance force during the flexion and extension processes. When the user bends forward, it
can provide lower resistance force for human bending motion by supplying a pressure of 500 kPa. This can overcome the shortcomings of passive power assist devices that just use elastic elements as actuators.

### 3.3 Layer-type pneumatic actuator

Actuator B is a layer-type pneumatic actuator composed of two TPU balloons. The TPU material is a composite material that combines the properties of rubber and plastic. It has excellent weight bearing capacity and impact resistance, and it is widely used in producing massage chairs and airbags. The TPU balloon used in this device is 150 mm long, 100 mm wide, and 2 mm thick, as shown in Fig. 17. When the balloon is supplied with compressed air, it will become taller, reaching 50 mm.

![Overview of TPU balloon](image)

**Fig. 17** Overview of TPU balloon

A TPU balloon can take a maximum air pressure of 250 kPa. Fig. 18 shows the relation between expansion force and displacement in height. The expansion force reaches 450 N at a pressure of 60 kPa.

![Relation between expansion force and displacement in height](image)

**Fig. 18** Relation between expansion force and displacement in height
TPU balloons are put inside pockets made with nylon bands. The complete structure of a layer-type pneumatic actuator is shown in Fig. 19. In this device, the actuator is installed in the inner layer of the garment. In order to increase the moment arm \( A \) of assistance force, the expansion displacement in height can be adjusted by changing the pressure applied.

![Fig. 19 Structure of layer-type pneumatic actuator](image)

### 3.4 Conclusion

In this chapter, according to the analysis of former chapter, the key point of designing a power assist wear is to choose the appropriate actuators to output the assistance force. At first, I choose McKibben-type pneumatic rubber artificial muscles as actuator A to provide the assistance force, and subsequently improved by using elongation type pneumatic rubber artificial muscles as actuator A. For this type of muscles have a larger contraction rate, its properties are more similar to biological muscles and is better to be used for low back support. The related experiments have proved its characteristics.
4 Wearable IMU sensor system

In this study, in order to provide the assistance force automatically, the system need to measure the body’s movement and then related sensor system is required.

4.1 Human motion detection

There are many methods to detect the human’s movement. In laboratory or hospital environment, the method of using high-speed cameras and reflective mark is commonly used. These systems are usually based on image analysis techniques to calculate the coordinates of the mark and then get the related angles of human body. These systems can not provide the real-time data, most of them are independent system, and can not be merged with other systems. By the way, these systems are used in a fixed location, can not be used in the outdoor environment. So they are not suitable to be used for wearable power assist device.

For the exoskeletal type power assist device, it is easy to install the encoder or potentiometer sensor on the link joint. But for the power assist wear like normal clothing, a new approach to measure the human motion is needed.

4.2 Wearable IMU sensor system

In recent years, more and more gyroscopes and accelerometer applied to Apple and Nintendo games devices, while its prices are declining, inertial sensor is getting better and better to fit for portable-use requirements. Many laboratories are using inertial sensors to detect acceleration and speed of human’s movement. A portable gait analysis system using gyroscope and accelerometer was developed by Department of Health Technology and Informatics, Hong Kong Polytechnic University, this study explores the use of support vector machine (SVMs) to classify different walking conditions for hemi paretic subjects[32]. A wearable gait analysis system using angular rate sensor and acceleration sensor was developed by TADANO Lab, Hokkaido University, angular rate sensor and acceleration sensor are attached to a human leg to calculate the joint position and joint angle data according to the leg angular acceleration during walking[33]. A wearable sensor system for lower limb joint angle measurement was developed by WATANABE Lab, Tohoku University, using gyroscope and accelerometer two components, this system can correct gyroscope signals from accelerometer signals in measurement of hip, knee and ankle joint angles with a wireless wearable sensor system[34]. The Kalman filter is useful for combining data from several different indirect and noisy measurements[35]. It weights the sources of information
appropriately with knowledge about the signal characteristics based on their models to make the best use of all the data from each of the sensors. There is no such thing as a perfect measurement device; each type of sensor has its strong and weak points. The idea behind sensor fusion is that characteristics of one type of sensor are used to overcome the limitations of another sensor.

As it is illustrated in Fig. 20, sensor fusion algorithm is used that gyro sensor and accelerometers are combined together to obtain stabilized angles[36]. The gyro sensor is used as the underlying measurement of angular velocity. It is used to help the accelerometer separate out gravitational and linear acceleration components, and by integrating the output of gyro sensor in real time to find the raw angle data.

![Sensor fusion algorithm](image)

In this research, in order to measure human motion a 5-DOF Inertial Measurement Unit (IMU) composed of two gyro sensors AE-GYRO-SMD (ENC-03R; Murata) and a triple-axis accelerometer (KXP84-2050; Kionix) are used, which is shown in Fig. 21. The orientation of axes of sensitivity and polarity of rotation are defined as in Fig. 22.

![IMU components](image)

(a) AE-GYRO-SMD (ENC-03R; Murata)  
(b) AE-KXP84 (KXP84-2050; Kionix)
(c) 5-DOF IMU sensor

Fig. 21  IMU sensor composed of gyro sensors and accelerometer units

Fig. 22  Orientation of axes of sensitivity and polarity of rotation

Fig. 23  Definition in terms of human joints
Three IMU sensors are mounted on a human subject's chest, upper leg, and lower leg, as shown in Fig. 23. Using IMU sensors, the pitch angles of the trunk, upper leg, and lower leg can be measured and then calculate the related angles of human joints, which are defined as $\theta_{ht}$, $\theta_{hk}$, and $\theta_{ha}$, respectively. $\theta_{tr}$ is defined as the angle of trunk flexion; it can be calculated using Eq. (13).

$$\theta_{tr} = \theta_{ht} + \theta_{hk} + \theta_{ha} = 90^\circ$$ (13)

### 4.3 Experiment in Quick MAG system

In order to verify the precision of wearable IMU sensor system for posture detection, an autotracking video-recording system (OKK, Quick-Mag, Japan) is used from the left side in the sagittal plane. Markers are placed on the left side to delimit the body segments which is shown in Fig. 24.

![Fig. 24 Experiment in Quick MAG system](image)

Quick MAG system use high-speed camera to capture the markers attached to human body, get the three-dimensional coordinates of the markers by using image processing technology. And a five-segment model is used to calculate the joint moments around the hip, knee, and ankle joints, and then get the angles of different joint which are defined in Fig. 25[37].

The related angle between human trunk, upper leg and lower leg are shown in Fig. 26.
From the above curve data, it can be seen that wearable IMU sensor system can detect the
body's movements in real time mode. For the difference between two curves, there may be two reasons: First, it is induced by muscle movement because the IMU sensor unit is attached to human body closely. Second, spine is not actually rigid structures, human spine cannot be seen as a rigid rod structure when human upper body is bending forward; there is an adaptive change in order to improve the stability of spine. But in Quick MAG system, human spine is just considered as a rigid rod using two marks.

4.4 Conclusion

In this chapter, I talk about the wearable Inertial Measurement Unit (IMU) sensor system. For measuring the human’s motion, inertial sensors are selected based sensor fusion technology by combine accelerometer sensors and gyro sensors. In order to verify the precision of wearable IMU sensor system for posture detection, I use Quick MAG system to detect human body's posture and compare the difference of two methods.
5 Evaluation of assistance effectiveness

For the power assist wear proposed in this study, appropriate control strategies should be selected, and the effectiveness should be verified through experiments.

5.1 Device control strategy

In order to provide efficient assistance force, it is important to develop appropriate control strategies to minimize the interference of human actions. Human erector spinae muscles and rectus abdominis muscles work together for the balance required to maintain a person's posture. Therefore, as a human bends forward, rectus abdominis muscles need to overcome a larger resistance force when support is applied to the back muscles for the extension motion.

A related experiment was done to choose an appropriate control method. In the experiment, the subject was told to perform flexion and extension motions by supplying elongation-type pneumatic rubber artificial muscle with the appropriate different pressure, where $P_5$ is the setting value of the elongation-type pneumatic rubber artificial muscle. The layer-type pneumatic actuator was supplied with constant pressure and then sealed during the experiment. Here I can take this actuator as a sensor and measure its response. The pressure of layer-type pneumatic actuator $P_2$ and the angle of trunk $\theta_f$ are illustrated in Fig. 27.

![Diagram of force and pressure](image)

(a) $\Delta F_b = \Delta P_2 \times S$

(b) $P_2=0$ kPa  

(c) $P_2=500$ kPa

Fig. 27  Layer-type pneumatic actuator pressure response during flexion motion
In Fig. 27, when the pressure of the elongation-type pneumatic rubber artificial muscle $P_e$ is set to 0 kPa, the output force $F_y$ is the largest. The subject therefore needs to overcome the greatest resistance to a forward bend; the back needs to bear a large force $AF_B$ for the bending motion. The value of $AF_B$ has a linear relationship with $AP_2$ and $S$, where $AP_2$ is the relative change of pressure $P_2$ and $S$ is the contact area between layer-type pneumatic actuator and the human back.

The following control strategy is proposed for the power assist wear. During the bending forward process, the pressure of actuator A, $P_e$ is set to 500 kPa to minimize the resistance force for the rectus abdominis muscles. When assistance force is needed in order to lift a load using extension motion, $P_e$ is set to 0 kPa to provide power assistance support. For actuator B, consider of the required moment arm and the comfort of human body, the pressure of actuator B is set to a constant value no more than 30 kPa[38].

---

**Fig. 28** Structure of control system

**Fig. 29** Block diagram of control system
The whole structure of the control system is shown in Fig. 28. The system consists of A/D board (PCI-3133, Interface Corporation), D/A board (PCI-3341A, Interface Corporation), pressure sensor (AP-43, KEYENCE Corporation), pneumatic servo valve (EVD-1500-008AN, CKD Corporation), and IMU sensors.

The block diagram of control system is shown in Fig. 29. Through the IMU sensor mounted on the human body, the sequence of human movement can be determined. At the time that the power assistance is needed during the human bends forward and lifts the load, the reference pressure $P_s$ is determined through the posture to pressure algorithm, the controller will output the desired driving voltage on the pneumatic servo valve, regulate compressed air, and then the required assistance force can be obtained. As for the posture to pressure algorithm, it will be described in Section 5.4.

5.2 Evaluation of assistance effectiveness using sEMG signals

The power assist wear proposed in this paper is driven by elongation-type pneumatic rubber artificial muscles set parallel to the erector spinae muscles, so it is essential that the sEMG activity of erector spinae muscles be measured in order to verify the effectiveness of the assistance.

The sEMG signals of erector spinae muscles are measured using wireless EMG instruments (Km-818MT; Mediarea support business union). The default sampling period is 1ms. In order to evaluate the assistance effectiveness, iEMG (integral of sEMG) is introduced as Eq. (14).

$$iEMG(t) = \sum_{i=t-N+1}^{t} sEMG[i]$$

(14)

where the epoch of length N is 250.

![Fig. 30 Lifting load experiment](image)

In the experiments, sEMG electrodes were pasted on the both sides of the L5/S1 joint to measure the activity of the erector spinae muscles, as shown in Fig. 30. Because the
layer-type actuator was installed just on the top of the electrodes, in order to keep the electrodes from being separated from the skin surface, I used a lid to cover them. The subject was told to perform the following actions:

From 0 seconds to 4 seconds, stand upright on the ground.
After 4 seconds, begin to bend forward until a 60-degree angle is reached, and then maintain the same posture until second 18.
After 18 seconds, begin to straighten until the upright posture is restored.

All the movements were performed both with and without power assistance, and related sEMG data was transformed into iEMG using Eq. (14). The iEMG data and the angle of the trunk \( \theta_T \) in the two experiments are illustrated in Fig. 31 and Fig. 32. The pressure of the elongation-type pneumatic rubber artificial muscle is shown in the following figures, where \( P_s \) is the setting value of the artificial muscles and \( P_m \) is the measured value. Here, the pressure of the layer-type pneumatic actuator was set to a constant value of 15 kPa.

![Graphs showing iEMG and angle of trunk](image)

**Fig. 31** Evaluation by measuring sEMG signal of erector spinae muscles (loads: 0 kg)

In the first experiment, the subject just maintained a bent forward position without lifting a
load. While the subject was bent forward, since the erector spinae muscles participate in maintaining the balance of posture, the iEMG signal increased slightly. In accordance with the control strategy, before extension, pressure $P_s$ was set from 500 kPa to 0 kPa to provide the largest assistance force when the subject was wearing the power assist wear. Because of the assistance force, the iEMG signal was lower than it was without assistance when the upper body was being restored to the upright position, as indicated by blue dotted lines in Fig. 31. The reduction in iEMG signal maximum value was about 19%.

![Graph showing iEMG and $\theta_t$ signals without assistance](image1)

(a) Without assistance

![Graph showing iEMG and $\theta_t$, $P_s$, and $P_n$ signals with assistance](image2)

(b) With assistance ($P_s=500$ kPa $\rightarrow$ 0 kPa)

**Fig. 32** Evaluation by measuring sEMG signal of erector spinae muscles (loads: 12.6 kg)

In the second experiment, the subject was instructed to bend forward, pick up a heavy load (12.6 kg) from the ground, and then return to an upright position while holding the load. According to the control strategy, before picking up the load, the pressure $P_s$ was changed from 500 kPa to 0 kPa to provide the largest assistance force with the power assist wear in place. The iEMG signal was significantly lower with assistance than without at two particular times, as indicated by blue dotted lines in Fig. 32. The first was at about second 15, when the heavy load was picked up; the second was at about second 18, when the upright posture was restored. Because the erector spinae muscles were supported by a related assistance force, the reductions in iEMG signal maximum value at the two times were about 38% and 29%,

- 31 -
respectively.

5.3 Floor to waist lifting test on PrimusRs

Fig. 33 Floor to waist lifting test on PrimusRS

(a) Without assist

(b) With assist

(c) Output power

Fig. 34 Maximum lifting weight and output power
As illustrated in Fig. 33, floor to waist lifting test is conducted in order to determine the subject's lifting capacity using PrimusRs (BTE Technologies, USA) when wearing the power assist wear, this device can provide concentric/eccentric resistance through an entire range of motion on a safe and dependable basis. Lifting is conducted with an initial weight of 100.0 N and is terminated at a safe maximum lifting weight which is illustrated in Fig. 34.

As shown in Fig. 34, without assist the maximum load that the subject can lift is 35 kg, and the speed is gradually reduced as the load is setting as more larger. But with assist the speed remains almost the same until to 35 kg, the maximum lifting capacity can reach to 40 kg, and the related output power of with assist is significantly larger than without assist.

5.4 Improvement on control strategy

In the Section 5.3, the system just uses ON/OFF signal to control the air pressure valves. For practical use, it will be inconvenient for the wearer to operate a switch when the desired assistance is needed. The system should be able to provide the assistance force automatically.

![Graphs showing force and angle over time](attachment:image.png)

(a) Stoop lifting mode  (b) Squat lifting mode

Fig. 35  Pitch angles of body during lifting cycle

In order to provide the desired assistance force automatically, the system needs to be able to know the sequence of the human's action and the characteristics during the upper body is
bending forward and lifting the load. For lifting activities, human usually use two modes: squat lift or stoop lift. For squat mode, human prefer to bend his knees; for stoop mode, human prefer to bend his back. Which one is better when lifting load, it is an ergonomic debate that has not been solved. In this study, related experiments were conducted for measuring the characteristics of human lifting action by using a force plate (ToMoCo-FPm) in order to find a better control strategy.

During the experiments, the subject (weight: 70 kg) is told to stand on the force plate and lift a load (12.6 kg) from the ground in different mode three times in 25 seconds. By using a force plate and wearable IMU sensor system, the pressure force on the z-axis and the angles of the body (θ_r, θ_k, θ_a, θ_d) are recorded at the same time, and related data are illustrated in Fig. 35.

As shown in Fig. 35(a), for stoop mode, the maximum angle of trunk θ_r is about 90 degrees; the angle of knee θ_k is always maintained as about 180 degrees; the angle of ankle θ_d is larger than 90 degrees in order to balance the body when bending forward and then lifting a load. In Fig. 35(b), for squat mode, the maximum angle of trunk θ_r is about 65 degrees, is smaller than in stoop mode; the angle of knee θ_k is varied from 70 degrees to 180 degrees, and is not maintained at a same value; the angle of ankle θ_d is smaller than 90 degrees in order to balance the body when bending forward and then lifting a load.

Due to the subject’s movement, dynamic changes on the force plate was occurred, especially at the time when the subject was lifting the load, the maximum values of the force in stoop mode is about 900N, while the maximum values of the force in squat mode is about 1000N, and is larger than in stoop mode when lifting the same load. Therefore, it is not the reliable and adequate method to detect the changes of the load just using the force plate.

But it can be found that the most appropriate time to provide the assistance is that when the value of the force is becoming the largest, the related angle of human trunk θ_r also reaches to the maximum values. For the angle of trunk is calculated using Eq. (13), so using the related three angles, the best time to provide the assistance force can be gotten.

As shown in red dotted line in Fig. 35, for different lifting mode the best time to provide the assistance force is when the three angles become the extreme values. So it will be possible in the future that the system can just use wearable IMU sensors to control the power assist wear to output the required assistance automatically. The proposed angle to pressure algorithm in one cycle is illustrated in Fig. 36.
Fig. 36  The angle to pressure algorithm in one cycle
5.5 Conclusion

In this chapter, the effectiveness of the device have been verified through related experiments. First the appropriate control strategies were determined to minimize the interference of human actions; for the power assist wear is driven by elongation-type pneumatic rubber artificial muscles which is set parallel to the erector spinae muscles, so the sEMG activity of erector spinae muscles were measured in order to verify the effectiveness of the assistance; in order to determine the wearer’s lifting capacity when wearing the power assist wear, related floor to waist lifting test was conducted on PrimusRs (BTE Technologies, USA) test platform; in order to provide the desired assistance force automatically, related experiments were conducted for measuring the characteristics of human lifting action by using a force plate (ToMoCo-FPm).
6 Static holding test

For the power assist wear proposed in this study, the elongation-type pneumatic rubber artificial muscle can be seen as a spring with a variable stiffness. Compared with exoskeleton type power assist device, which has the advantage during bending forward and maintaining a static posture.

6.1 Static holding test

Related research suggests that antagonist muscle co-activation is necessary for aiding ligaments in maintaining joint stability during loaded tasks, for the human spine, its stability requires agonist-antagonist co-activation(Fig. 37)[39]. A significant reduction in the activity of back muscles had been found due to wearing a lumbosacral orthosis (LSO) during an unstable sitting task[40][41]. When the power assist wear proposed in this study is worn on the human body, the similar effect should be get like LSO to maintain a static posture.

![Image of spine and device]

Fig. 37 Spine stability requires agonist-antagonist co-activation

In order to assessment the stability of human body when wearing the device, static holding test is conducted on force plate to measure the body’s movement.

During the experiment, the subject is told to stand on a force plate, bend forward, and maintain the static posture with holding a load (7 kg) in both hands during 30 seconds, as illustrated in Fig. 38. The force plate recorded the movement of the COG.
The system default sampling frequency is 60 Hz, 1800 coordinate values in the X-axis and the Y-axis were obtained respectively during the experiment. In order to evaluate the movement of the COG, the collected data is analyzed according to the following equations:

$$\Delta L_i = \sqrt{(X_{i+1} - X_i)^2 + (Y_{i+1} - Y_i)^2}$$  \hspace{1cm} (15)

$$LNG = \sum_{j=1}^{n} \Delta L_j$$  \hspace{1cm} (16)

$$LNG_x = \sum_{j=1}^{n} |X_{i+1} - X_i|$$  \hspace{1cm} (17)

$$LNG_y = \sum_{j=1}^{n} |Y_{i+1} - Y_i|$$  \hspace{1cm} (18)

$$VX_{i+1} = (X_{i+1} - X_i) / T$$  \hspace{1cm} (19)

$$VY_{i+1} = (Y_{i+1} - Y_i) / T$$  \hspace{1cm} (20)

where the $\Delta L_i$ is the moving length of the COG in one sampling period, $LNG$ is the whole moving length in 30 seconds, and n is the total number of samples (n=1800). $LNG_x$ and $LNG_y$ are the related moving length in the direction of the X-axis and the Y-axis. $VX_{i+1}$ and $VY_{i+1}$ are the related moving velocity in the direction of the X-axis and the Y-axis, here $T$ is the sampling period ($T=1/60$ s).

### 6.2 Test result

The moving length of the COG are calculated and shown in the Tab. 2. Related data are
shown in Fig. 39, the front two figures are the moving trajectory of the COG, the last two figures are the velocity distribution of the COG.

Fig. 39 The moving length and velocity distribution of the COG

<table>
<thead>
<tr>
<th>Tab. 2</th>
<th>The moving length of the COG</th>
</tr>
</thead>
<tbody>
<tr>
<td>Unit: m</td>
<td>With assistance</td>
</tr>
<tr>
<td>$LNG_x$</td>
<td>0.1919</td>
</tr>
<tr>
<td>$LNG_y$</td>
<td>0.332</td>
</tr>
<tr>
<td>$LNG$</td>
<td>0.4186</td>
</tr>
</tbody>
</table>

It is obviously that with assistance the moving length of the COG is smaller than without assistance situation, the corresponding velocity distribution of the COG tends to center
distribution. This data indicates that the stability of human body is improved for the assistance during static holding situation. So the power assist wear will be effective for the nursing caregivers who need bending their bodies during working times.

6.3 Conclusion

In Chapter 6, in order to get the assessment of the stability of human action when wearing the device, static holding test was conducted on force plate to measure the body’s movement of the center of gravity (COG), the result of experiment has shown that this device can be effective for human stability. This power assist wear can reduce the incidence of the LBP, but also can improve the wearer's stability.
7 Conclusion and future developments

In this study, I have proposed a power assist wear for low back support, a device using new types of pneumatic actuators. Compared to that of McKibben-type pneumatic rubber artificial muscle, the contraction rate of elongation-type pneumatic rubber artificial muscle is larger. As it does not use an exoskeleton structure, the device can be worn on the human body just like normal clothing. It can provide assistance force for the low back, reducing the possibility of LBP, and the assistance power of the device can be adjusted by changing the pressure of the compressed air. The effectiveness of the device has been verified through experiments.

Chapter 1 talks about aging society’s needs of power assist robotic technology, and related research in the world. Based on this demand for wearable assist robot, according the idea of designing power assist device like normal clothing, a new wearable power assist wear is proposed by using pneumatic actuators. The principles of designing a wearable power assist wear are determined: safety, user-friendliness, convenient to operate, and has a low costs.

In Chapter 2, the biomechanical model of the human spine is analyzed, and get to know the main reason of LBP: when the human is bending forward and lifting a load, for the erector spinae muscles have a small lever arm, the spine has to bear a large amount of force, which is several times of body weight. Once the disc compression force and shear force are over the limited scope that the spine can bear, permanent damage will result and lead to LBP. In order to provide the power assistance, it can be done to support the human’s back from two aspects: increasing the assistance force and related lever arm length. This is determined as the configuration of device and its operation: Using pneumatic rubber artificial muscle which is fixed parallely to the erector spinae muscles, it can provide external assistance force, and will increase the effective amount of torque; Using layer type pneumatic actuator, it can increase the lever arm length of assistance force; At the same time, due to the expansion, it will tighten the garment and then increase the intra-abdominal pressure (IAP), which can lead to increased stability of the spine.

In Chapter 3, according to the analyzation of former chapter, the key point of designing a power assist wear is to choose the appropriate actuators to output the assistance force. At first, McKibben-type pneumatic rubber artificial muscles were chosen as actuator A to provide the assistance force, and subsequently it is improved by using elongation type pneumatic rubber artificial muscles as actuator A. For this type of muscles have a larger contraction rate, its properties are more similar to biological muscles and is better to be used for low back support. The related experiments have proved its characteristics.
In Chapter 4, I talk about the wearable Inertial Measurement Unit (IMU) sensor system. For measuring the human’s motion, inertial sensors are selected based on sensor fusion technology by combining accelerometer sensors and gyro sensors. In order to verify the precision of wearable IMU sensor system for posture detection, Quick MAG system is selected to detect human body's posture and compare the difference of two methods.

In Chapter 5, the effectiveness of the device have been verified through related experiments. First, the appropriate control strategies were determined to minimize the interference of human actions; for the power assist wear is driven by elongation-type pneumatic rubber artificial muscles which is set parallel to the erector spinae muscles, so the sEMG activity of erector spinae muscles were measured in order to verify the effectiveness of the assistance; in order to determine the wearer’s lifting capacity, related floor to waist lifting test was conducted on PrimusRs (BTE Technologies, USA) test platform; in order to provide the desired assistance force efficiently, related experiments were conducted for measuring the characteristics of human lifting action by using a force plate (ToMoCo-FPm).

In Chapter 6, in order to get the assessment of the stability of human action when wearing the device, static holding test was conducted on force plate to measuring the body’s movement of the center of gravity (COG), the result of experiment has shown that this device can be effective for human stability. This power assist wear can reduce the incidence of the LBP, but also can improve the wearer's stability.

At last, in order to improve the performance of the device for practical using, further improvements are necessary. Such as the integration of the actuators, will make it really like the normal clothing, and makes it easier to wear; The improvement on decreasing the power consumption of the whole system that the system uses 15 A/D acquisition channels to measure the analog signals now. With the development of MEMS technology, there are some IMU chips that can output the signal into digital. In order to realize the system integration and miniaturization for wearable using, smaller control unit is also needed to fulfill these requirement.
REFERENCES


[30] Ching-Ping Chou and Blake Hannaford. Static and dynamic characteristics of McKibben pneumatic


Publication

[1. Reference Papers]

Xiangpan Li, Toshiro Noritsugu, Masahiro Takaiwa, and Daisuke Sasaki.

[2. OTHER PAPERS, ETC.]

(1) Other Papers


(2) ORAL AND POSTER PRESENTATION

① Research on Gait Rehabilitation Training Robot Driven By Pneumatic Actuators. Han Jianhai, Li Xiangpan, Zhao Shushang. Proceeding of the ICFP2009, (The 7th International Conference on Fluid Power Transmission and Control), April 7-10,(2009), 570-573.

② Design of wearable power-assist-wear for lower back support using pneumatic actuator, 空気圧アクチュエータを用いた腰部パワーアシストウェアの開発, Xiangpan Li, Toshiro Noritsugu, Masahiro Takaiwa, and Daisuke Sasaki. 第 30 回日本ロボット学会学術講演会予稿集(CD-ROM), No.815 (2012 年 3 月)

③ Design of wearable power assist wear for low back support, 腰部支援用エアラブルパワーアシスト装置の設計. Xiangpan Li, Toshiro Noritsugu, Masahiro Takaiwa, and Daisuke Sasaki. 第 30 回日本ロボット学会学術講演会予稿集(CD-ROM), No.2D2-5 (2012 年 9 月)

④ Development of power assist wear for low back support using pneumatic actuators. Xiangpan Li, Toshiro Noritsugu, Masahiro Takaiwa, and Daisuke Sasaki. The Eighth International Conference on Fluid Power Transmission and Control (ICFP 2013), Hangzhou, 2013, April 9-11. (Accepted)

(3) PATENT

Pneumatic driving type exoskeleton mechanical structure of lower limb walking rehabilitation training Robot 韩建海, 赵书尚, 郭冰青, 李向龙, 程卫卫. CN 201870775 U (22-Jun-2011) Applicants: UNIV HENAN SCIENCE & TECH
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