Lifting heavy objects increases the risk of low back pain. We calculate the load exerted on the human body during an actual lifting operation. To assist lifting, we propose an endoskeleton-like suit that exerts an assistive force. The arms, waist and legs have pneumatic rotary actuators driven directly by micro air pumps supplied by portable Ni-Cd batteries. The muscle forces are sensed by a new muscle hardness sensor utilizing a sensing tip mounted on a force sensing film device. The embedded microcomputer calculates the necessary joint torque for maintaining a position according to the equations derived from static body mechanics using the joint angles, and the necessary joint torque is combined with the output signals of the muscle sensors to make control signals. The suit was applied practically to a human body and movement experiments that weights in the arms were held and take up and down was performed. Each unit of the suit could transmit assisting torque directly to each joint verifying its practicability.

Introduction

The development of welfare machines which can meet the requirement of the elderly is now an important subject due to the rapidly aging population. The research for the robot which was aimed at supporting transfer movements of patients was begun in the 1970s, and MEL-Kong was the representative example, and an operation robot Nurcy based on the master and slave control system was developed. However, none of these show any prospects of the utility yet.

In recent years, the developmental research involved in solving this problem has been activated. The device which can support the transfer movements of patients and the device which can support the movements of carrying a patient in one’s arms are proposed. The devices use the electric motor and gear as an actuator.

In order to develop an wearable power assisting suit which gives nurses the extra muscle they need to lift their patients, we fabricated powered arms using pneumatic rubber tube actuators in 1991, and then formed the waist and legs constructing a suit to be worn by a nurse.

Then, we developed a stand alone wearable power assisting suit by a substantial miniaturization of the power supply and control systems using micro air pumps, portable Ni-Cd batteries, and an embedded microcomputer. In
In this paper we give the characteristics of a new wearable power assisting suit and prove the possibility that the suit can be practically utilized.

**The basic design concepts**

There are four basic points in the design of the power-assist suit:

1. The system must be absolutely safe, i.e. ready for all emergencies. This is assured by placing the nurse in control, i.e. the assist system is a master and slave system in one unit. In addition, when the electric power source is cut off, the air supply pumps and the exhaust valves prevent back flows from the air actuators, so that the suit continues to assist in the process of holding.

2. There are no mechanical parts on the front of the suit. The nurse’s arms and chest may be in direct contact with the body of the patient carried in her arms. This produces empathy between the patient and nurse.

3. Flexible joints are implemented by a pneumatic rotary actuator using rubber cuffs. The use of these pneumatic actuators in joints make the nurse’s arms, waist and legs soft to touch.

4. Assisting forces adapted to requirements for bending and stretching the joints is achieved by using the muscle hardness sensor to detect the force exerted in the muscles driving the joints. Thanks to this sensing system, smooth movements of the arms, waist and legs of the assisting suit are possible. As an additional backup and failsafe mechanism, the joint torque needed to maintain a position is calculated by means of equations derived from static body mechanics using the joint angles.

The photograph of the stand alone type wearable power assisting suit and construction are shown in Fig. 1. The shoulder of the arm unit can swing back and forth and side to side. The joints of the suit have double axles so that the each unit can bend with the bending of the arm, waist and leg. The joints of the elbows, waist and knees are rotated by newly developed direct drive pneumatic rotary actuators which are driven by micro air pumps applied by portable Ni-Cd batteries. An embedded microcomputer and PWM driving circuits are mounted on the back. The portable Ni-Cd batteries are attached to the legs. These units are fabricated of duralumin alloy. The weight of the suit is about 30 [kg]. When the wearer stands upright, the entire weight of the suit can be supported by the leg units, and when the wearer bends at the waist or knee, the weights of the waist and arm units can be supported by the actuators.
The sensing and control systems of the power assisting suit are shown in Fig.2. The exerting muscle forces of the arms, waist and legs of the nurse are detected by the muscle hardness sensors placed on the nurse’s upper arms (biceps brachii muscle), on the leg!; above the knees (rectus femoris muscle) and on the back above the hip (erector spinae muscle). The output signals of the sensors are transmitted to the embedded microcomputer. The embedded microcomputer calculates the necessary joint torque for maintaining a position, and the necessary joint torque is combined with the output signals of the muscle sensors to make control signals inputted into the PWM driving circuits. Then the supply of air flow to the cuff changes in accordance with the necessary joint torque.

**Controller**

SOPC (System on Programmable Chip) technology was used to implement the controller of powered assisting suit. The controller consists of an APEX20K200E (200K gate) FPGA device, A/D converters, ethernet controller, two external SRAMs and EEPROM memories on a single board. This FPGA device board has 9x12 [cm] width and 2.5 [cm] height. The controller core is a 32-bit wide Nios 2.0 processor and control block module. The control block contains 24 PWM channels (18bit), a interface logic for AID, FIR (Finite Impulse Response) filters and PID core (16 bit). The control block is able to operate even if alone and the calculation delay of the control block is only 20 clocks (The controller runs at a clock speed of 33MHz). The controller hardware for power assisting suit uses approximately 85% of the FPGA device.
Body Mechanics and calculation of necessary joint torques

First, we must determine the assistive force provided by the assist suit. To this end, we estimated the torque of the waist joint that induces low back pain during movement, and set this torque as the target.

A. Calculation method

The behavior of the wearer is modeled on a manufacturing plant worker (see Fig.5) and is estimated as follows. The worker undertakes lifting and unloading motions that are expected to induce back pain.

The human body is regarded as a linked system with five degrees of freedom, as shown in Fig. 6. Gauge marks are attached to the relevant joints. The angle of each joint and the acceleration of the object are obtained by measuring a target point through video analysis software Movias. The measured values are substituted into Equation (3), which determines the torque $T$ of the waist joint. The model parameters are listed in Table 1. The height and weight of a subject is assumed as 1.65 m and 70 kg, respectively. Centroid distances and weights of the various body parts are assumed as their average values.
B. Calculation results

The torque dynamics of the waist joint, calculated from the measured values, are plotted in Fig. 3. The torque is maximized at the moment of lifting the nearest object from the ground. The motion of lifting this object is highlighted by the colored rectangle in Fig. 7. The left and right edges of this rectangle denote the start and end times of the lifting motion, respectively. The torque generated during lifting is estimated to reach up to 150 Nm. Therefore, the assist suit design was based on this value.

\[
T_1 = (M' L_4 a + M_1 L_{q1} g) \cos \sum_{i=2}^{5} \theta_i
\]

\[
T_2 = (M_2 L_{q2} + (M' + M_1) L_2) g \cos \sum_{i=2}^{5} \theta_i + T_1
\]

\[
T = (M_3 L_{q3} + (M' + \sum_{i=2}^{5} \theta_i) L_3) g \cos \left( \sum_{i=2}^{5} \theta_i \right) + T_2
\]

**Measurement of muscle force**

New muscle sensors were developed for measuring the exerted muscle force which could apply to various conditions of subcutaneous fat. As shown in Fig.8, new muscle sensors consist of a contact projection formed of silicon rubber and aluminum on a force sensing film device, and the projection is held against the muscle. The optimal shape of the projection fix thin subcutaneous fat is cylinder shape and that for

<table>
<thead>
<tr>
<th>TABLE I Parameters of the motion model</th>
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<tr>
<td>( T_1 ) Torque of joint (Nm)</td>
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<tr>
<td>( M_i ) Mass of link i [kg]</td>
</tr>
<tr>
<td>( L_i ) Length of the link i [m]</td>
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<tr>
<td>( L_{q1} ) Centroid distance of link i [m]</td>
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thick one is cone shape. The arm muscle sensors are applied to the wearer's upper arms (biceps brachii muscle). The waist muscle sensors are applied to the wearer's back above the hip (erector spinae muscle). The two leg muscle sensors are applied to the wearer's legs above the knees (rectus femoris muscle) and counter side of this sensor. The characteristics of sensors and its experimental results are described in next sections.

![Fig.8 Muscle hardness sensors](image)

**Pneumatic Rotary Actuators**

The pneumatic rotary actuators of elbow, knee and waist joints are shown in Fig.9. These rotary actuators are constructed of pressure cuffs (90mmx120mm) of a sphygmomanometer sandwiched between thin plates. The thin plates are connected at one end in the case of the actuator for the waist joint and are connected to each other at both ends in a zigzag manner. These pressure cuffs are driven directly by micro air pumps (30 [mmcp], 65 [mm] long) and the air is exhausted air by micro solenoid valves (20 [mmcp], 25 [mm] long). Two portable Ni-Cd batteries (12 [VI, 30 [mmcp], 300 [mm] long) are used for this operation.

![Fig.9 Pneumatic rotary actuators](image)

**Muscle hardness sensors**

The characteristics of the muscle hardness sensors applied to the muscles of the arm, waist and leg during the sequential movements of bending and stretching the arms, waist and legs are shown in Fig.10. The outputs of the muscle sensors are the repelling forces of muscles which were converted by using the output voltage-press back force characteristics of the sensors. The angle of zero degree corresponds to the full stretching of the arms, waist and legs. The loads were applied by weights in the arms. The output signals of the muscle sensor which is applied on the biceps brachii muscle were almost proportional to the bending angle of the elbow joint. They showed sensitivity to the weight load though hysteresis existed. On the other hand, the output signals of the sensor applied on the erector spinae muscle showed a parabolic relationship to the bending angle of the waist joint with hysteresis. The output signals showed sensitivity to the weight load. The output signals of the sensor applied on the rectus femoris muscle showed an almost linear relationship to the bending angle of the knee joint. The output signals showed sensitivity to the weight load. These characteristics showed...
the applicability of the muscle sensor, because the hysteresis was small and could be linearized by computer software.

**Rotary actuators**

Fig. 11 shows the output torque versus supply pressure characteristics of elbow, waist and knee actuators on the condition that these actuators were attached to their respective units. The output torque was measured by using a spring scale kept at a right angle to each unit’s member functioning as a moment arm. The joint angle of zero degree corresponds to the condition that the cuffs collapse. The relationship showed linearity with hysteresis.

The rotary actuators for elbow joint and knee joint had almost the same characteristics, i.e., the conversion ratios of these actuators depended strongly on the angles of the joints. This was due to the construction of the actuators, i.e., the thin plates were connected to each other at both ends in a zigzag manner, so that the plates extended in an arc. However, on the other hand, in the case of the actuator for the waist, the thin plates were connected at one end, so that the plates spread like a fun.
Operation characteristics of knee joint

Operation characteristics were measured practically by applying the power assisting suit to a human body and performing the holding a person, who has 60 [kg], up and down procedure, i.e., the simulated operation of holding up and down of a patient. The desired assisting torque of each unit was set as 50% of the calculated joint torque for keeping the weight with the additional torque corresponding to the output of the muscle hardness sensor. PI control was used to control the cuff pressure. The relationship between the supply pressure to the actuator, the sensed repelling force of the muscle and the rotational angle of the joints of the units were measured. Operation characteristics of the leg unit are shown in Fig. 12 as typical data of power assisting suit. Zero degree corresponds to the full stretching of the knee. Red line describes angle of knee, blue line describes sensed repelling force of knee muscles by muscle hardness sensor and orange line describes supplied pressure which is generated based on the desired pressure described as green line. The operator was lifting up a person after 7 [sec] from started time. This figure also describes that the necessary repelling force was generated effectively because that the repelling force was generated near time at the muscle becomes harder. Lack in the supply pressure at the knee stretching occurred because the air pump could not make up the shortage of air supply due to the suction effect caused by the forced inflation of the pressure cuff during the stretching.

DEVELOPMENT OF THE ASSIST SUIT

A. Structure of the assist suit

A prototype of the assist suit is shown in Fig. 13. The device was developed to generate the waist torque. The actuator is attached to the human body by a belt, which we designed as a full-harness safety belt. Two straight-fiber-type artificial muscles are mounted to the back parts, and an artificial muscle is attached to each femur. The assistive force to the waist is increased by an amplification mechanism. However, this mechanism cannot provide sufficient assist force when solely attached to the artificial muscle. Therefore, the amplification mechanism was installed in a full-harness-type safety belt and was designed to provide sufficient assisting force. The weight of the assist suit is 6.2 [kg].
The prototype requires air pressure for movement. A schematic of the assist suit is shown in Fig. 14. The assistive force increases during lifting operations when air pressure is applied to the back part of the artificial muscles. And, the artificial muscles position is changed when applying air pressure to amplification mechanism. This structure can increase the assist force. The device also permits rotational movement of the upper body around the waist. The assistive force is further increased by applying air pressure to the fixed artificial muscle attached to the thigh; this mechanism provides a stable force.

However, the nonlinear properties of this artificial muscle destabilize the control. We therefore linearize the muscle properties using a mechanical equilibrium model. The shape model of the artificial muscle is shown in Fig. 16. In this figure, $\Phi_0$ denotes the circumferential angle, which depends on the amount of contraction $x$. Table II lists the complete set of parameters related to the artificial muscle. From the mechanical equilibrium model developed in previous research, we can relate the contraction $x$, the contraction force $F_a$ and the pressure $P$, and hence derive the desired linearized values of the contraction $xd$ and

1) **Straight-fiber-type artificial muscle:** The actuator for the proposed device is a straight-fiber-type artificial muscle, and is overviewed in Fig. 15. The artificial muscle is composed of natural rubber latex and axially arranged carbon fibers. Under air pressure, the artificial muscle expands in the radial direction and contracts in the axial direction, generating a contraction force that serves as the actuating force. The artificial muscle has several desirable features; flexibility, high output, light weight, and high contraction capacity.
output contraction $x$. The contractile force of the artificial muscle is calculated by the following equations, which are based on the mechanical equilibrium model:

\[
\phi_0 = \frac{2\alpha_0 (\phi_0)^{0.5}}{(l_0 - x_0)^2 + \alpha_0^2 x_0 d_0} \quad (\alpha = 1.4) 
\]

\[
F_u(\phi_0, P_t) = \frac{P_s G_{3u}(\phi_0) - G_{1u}(\phi_0)}{G_{2u}(\phi_0)} \quad \text{(5)}
\]

\[
G_{3u}(\phi_0) = 4Kd_1 \left[ \tan \phi_0 \frac{d_1}{d_0} \right] \left[ \sin \phi_0 - \sin \phi_0 \cos \phi_0 \right] \phi_0 \quad \text{(6)}
\]

\[
G_{2u}(\phi_0) = \frac{M \tan \phi_0}{d_0 \sin \phi_0} \left[ \frac{d_1}{d_0} \right] \left[ \sin \phi_0 - \sin \phi_0 \cos \phi_0 \right] \phi_0 \quad \text{(7)}
\]

\[
G_{1u}(\phi_0) = \frac{4Kd_1}{d_0} \left[ \frac{d_1}{d_0} \right] \left[ \sin \phi_0 - \sin \phi_0 \cos \phi_0 \right] \phi_0 + 2K \frac{d_1}{d_0} \sin \phi_0 - \frac{Kd_0 M}{4n} \sin \phi_0 \quad \text{(8)}
\]

where $x_d$ is the amount of contraction, and $\alpha$ is a constant that approximately relates the amount of contraction to the diameter. $l_0$ and $d_0$ are the initial length and diameter, respectively, of the muscle, and $K$ and $t$ denote the stiffness and thickness, respectively, of the rubber. $M$ is the coefficient of the artificial muscle fiber, $n$ is the number of fibers and $b$ is the fiber width. The subscript $i$ distinguishes the different artificial muscles.

<table>
<thead>
<tr>
<th>Table II. Artificial Muscle Parameters</th>
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<tr>
<td>$P_i$</td>
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<tr>
<td>$F_i$</td>
</tr>
<tr>
<td>$x_{d}$</td>
</tr>
<tr>
<td>$l_{a}$</td>
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<tr>
<td>$d_{a}$</td>
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<td>$K_{a}$</td>
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Fig. 17 compares the artificial muscle properties obtained by the mechanical equilibrium model with the experimentally obtained values. The mechanical equilibrium model adequately reproduces the experimental values.
Assisting the operation of lifting a heavy object by the contraction force of the artificial muscle.

2) Amplification Mechanism: The amplification mechanism is fitted with a bellows cylinder and three plates connected to a shaft (Fig. 18). The bellows cylinder is an extendible actuator that exerts an airpressured expansion force.

Fig. 18 Amplification mechanism

Fig. 19 plots the basic properties of the bellows cylinder. The air pressure applied to the bellows cylinder is 0.3 MPa. The expansion force exerted by the bellows is approximated by Eq. (9).

Fig. 19 Characteristics of the bellows cylinder

\[ F_s = 0.0774x^2 + 36.55x + 4240 \quad (x = 0 - 65 \text{ mm}) \]  

MODELING AND SIMULATION

A. Model formula of assist suit

Fig. 20 is a model of the assist suit worn on the human body. When assisting a lifting movement, the operation of the pneumatic bellows cylinder and the artificial muscle must be appropriately timed. This section examines the effect of increasing the assistance mechanism. The appropriate timing of the air pressure is also investigated.

Fig. 20 Model of the proposed assist suit

The assist suit is operated when the waist angle reaches \( \theta \), calculated as follows:
The torque $T$ when wearing the power assist suit is calculated by Eq. (11).

$$\theta = \cos^{-1}\left\{\frac{\left(\frac{x_1^2 + (x_2 + x_3)^2 - (x_1 - x_3)^2}{2L_1(x_1 + h_2x_3)}\right)^2 + \frac{x}{2}}{L_1^2}\right\}$$

In the above equations, $l_1$ and $h$ denote the natural lengths of the bellows cylinder and artificial muscles, respectively, and $L_1$ and $L_2$ are the distances from the waist to the artificial muscle mounting positions. $F_1$ and $F_2$ are the contractile force of the artificial muscle and the expansion force of the bellows cylinder, calculated by Eqs. (5) and (9), respectively. The contractile force is determined from the air pressure $p_1$ and the contraction amount $\_l_1$. Similarly, the expansion force is determined from the amount of expansion $\Delta h$ and the air pressure $p_2$.

**B. Effect of Amplification Mechanism**

Fig. 21 plots the waist torque generated at the waist joint, calculated by Eq. (11). The force generated by the artificial muscle is constant and equal to 1000 N. The bellows cylinder expands from 0 to 65 mm. Clearly, the amplification mechanism increases the assist efficiency of the artificial muscle.

![Graph of assistive force generated by expansion of the bellows cylinder](image)

**Fig. 21 Assistive force generated by expansion of the bellows cylinder**

(Contractile force of artificial muscle = 1000 N)

**C. Simulation of the assist timing**

In this simulation, the operation timing is varied, and the torque is calculated from the model formula (Eq. (11)). To ensure proper operation of the method, the simulated results are compared with the measured values of Fig. 7. The assist force is changed by altering the timing of operating the bellows cylinder in the waist joint. The altered value is compared with the values calculated in Section 2. To generate the maximum torque of 150 Nm, a pressure of 0.2 MPa is applied to the artificial muscle (the pressure applied to the bellows cylinder is assumed as 0.3 MPa). Fig. 22 plots the results of changing the timing of air supply to the bellows cylinders. The artificial muscle operates at the instant of lifting by the human. The waist joint angle ($\theta_3$ in Fig. 6) is assumed to have moved through 0–90°. Fig. 22
plots the results of pressure application at 0%, 50%, and 100% contraction of the artificial muscle.

According to the simulation results, when air pressure is applied late to the bellows cylinder, the torque at the time of operation is not sufficiently generated until the waist joints reach about 65°. On the other hand, when air pressure is applied to the muscle and bellows at the same time, the torque is achieved at approximately 30° of the hip angle. Therefore, to generate the desired torque, the artificial muscles and the bellows cylinder should be simultaneously operated.

In the next section, we examine the assist effect on the human body under simultaneous operation of the muscles and bellows. In addition, the assist is switched by operator when the subject starts motion. The control system is used Simulink in MATLAB. And, it is controlled by dspace.
References


