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Stand Alone Wearable Power Assisting Suit
– Sensing and Control Systems –

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Abstract

A stand alone wearable power assisting suit was developed, which gives nurses the extra muscle they need to lift their patients and avoid back injuries. The suit consists of shoulders, arms, back, waist and legs units to be fitted on the nurse's body. 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.

1. 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 [1] was the representative example, and an operation robot Nurcy [2] 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 [3] which can support the transfer movements of patients and the device [4] which can support the movements of carrying a patient in one's arms are proposed. The devices are aimed at reducing the burden on the waist of the care person. These devices use the electric motor and gear as an actuator.

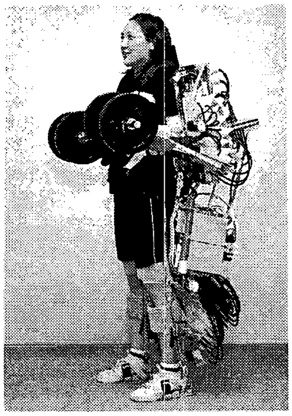
In order to develop an wearable power assisting suit which gives nurses the extra muscle they need to lift their patients, we fabricated powered arms using pneumatic rubber tube actuators in 1991 [5], and then formed the waist and legs constructing a suit to be worn by a nurse [6]. Then, we developed 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 [10]. 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.

2. Power Assisting Suit

The basic design concepts of the power assisting suit consists of four points, i.e., 1) an absolutely safe system. This is assured by the controllability by the nurse, i.e., the assisting system consists of a master and slave system in one unit. 2) The absence of mechanical parts in front of the suit. This results in good feelings between the patient and the nurse. 3) Flexible joints using pneumatic rotary actuators using pressure cuffs [8]. These joints give tender touch to the joints of the nurse. 4) Assisting forces correspond with the necessary forces to the joints. This is realized by developing muscle hardness sensor which detects the muscle force driving the joints [7]. In addition, as a backup and fail safe system, necessary joint torque for maintaining a position is calculated from static body mechanics.

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 [10].

A person who wearing the power assisting suit



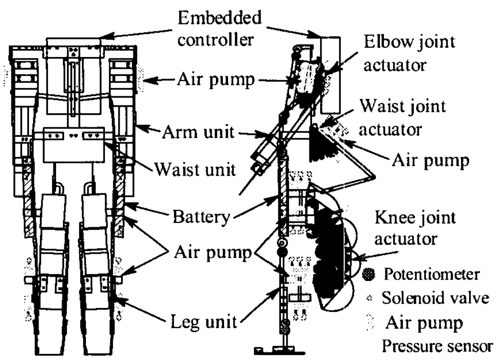


Fig.1 Power assisting suit

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 legs 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 [10].

3. 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 a 24 channel PWM (18 bit), a interface logic for A/D, FIR (Finite Impulse Response) filters and PID core (16 bit).

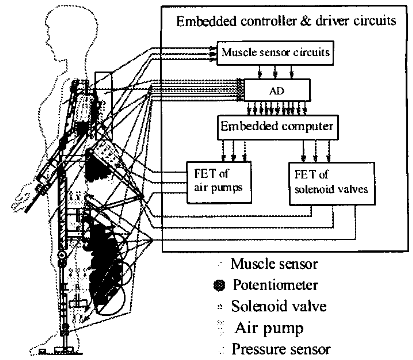


Fig.2 Sensing and control system of suit

A Block Diagram of the controller is shown in Fig.3. The control block contains 24 PWM channels (18bit), a interface logic for A/D, 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.

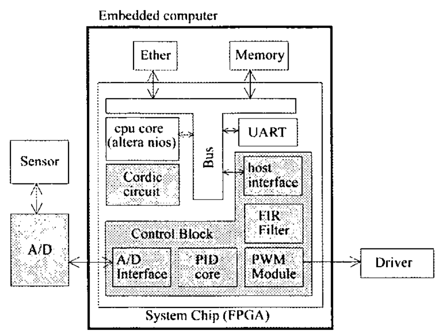


Fig.3 Microcomputer block diagram

We also developed the design tool to construct control blocks. In this development environment, the control block is composed of the combination of peripheral components (PWM, A/D IF, FIR, PID and so on). A graphical user interface (GUI) of the tool is shown in Fig.4. The design tool generates HDL (Hardware Description Language) codes of a control block and header file for user C language programs with some easy steps. Therefore, when users make the control block, a lot of knowledge is not needed.

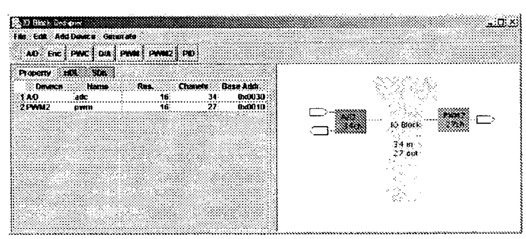


Fig.4 IO-block development tool

4. Body Mechanics and calculation of necessary joint torques

Necessary joint torque for maintaining a position of lifting a patient in her arm is calculated according to the equations derived from simplified static body mechanics as shown in Fig.5. In this model, three typical joint torques are taken into consideration. i.e., the elbow joint torque works as a substitute of the biceps brachii muscle, the waist joint torque works as an erector spinae muscle and the knee joint torque works as a rectus femoris muscle. The muscle forces driving the joints are estimated by computing the equations of moment balance, as a function of the joint angles. These equations of moment balance are derived from gravity compensation as the results of dynamics calculation using Lagrange's method.

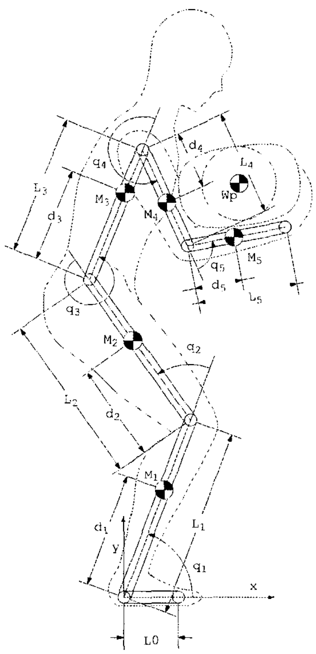


Fig.5 Body mechanics and link coordinate

Fig.5 also shows link coordinates of the power assisting suit. The subscript i indicates a number of the links in the following discussion. The lengths of the links are indicated by the symbol L_i, M_i denotes mass of links, q_i denotes the angle of joints and Wp denotes weight of the patience. The actual values of these parameters are shown in table 1.

link length	value [m]	weight of link	value [kg]
L_0	0.150	M_0	1.200
L_1	0.390	M_1	2.550
L_3	0.390	M_2	2.550
L_4	0.375	M_3	27.90
L_5	0.270	M_4	2.350
L_6	0.230	M_5	2.000

Table.1 Link parameters

For the simplicity, we assumed that the power assisting suits is symmetrical against a front face of this suit. Consequently we calculate equations as all joints are connected serially on lateral plain as shown in the Fig.5. Each center of gravity is indicated by marks which are composed of colors of black and white. The first (L_0), second (L_1), fifth (L_4) and sixth (L_5) links have the center of gravity onto its links. However, in fact, the third (L_2) and fourth (L_3) link has offset from center of gravity to the corresponding link, because that the third link is connected with pneumatic actuators and the fourth link is connected with pneumatic actuators and microprocessor unit. In addition, the sixth link must support forearm of nurse (it is assumed 1.1 [kg]). For theses reasons, the masses of the center of gravity are calculated in consideration of each unit's weight. Next table 2 shows the weight which must be supported by each links when the patient weight Wp is assumed as 60 [kg].

link	weight of each link [kg]
L_1	10.050
L_2	0.2675
L_3	37.900
L_4	2.3500
L_5	63.100

Table.2 Weight of each link

At first, a two dimensional homogenous coordinate transformation matrix is defined as follows:

$$T_i = \begin{bmatrix} \cos(q_i) & -\sin(q_i) & L_i \\ \sin(q_i) & \cos(q_i) & 0 \\ 0 & 0 & 1 \end{bmatrix},$$

where q_i is given as generalized coordinate. The posture of power assisting suit is calculated by the iteration of products of these matrixes and the equations of postures are represented as a function of joint angles q_i.

The gravity compensation does not use kinetic energy to calculate Lagrange's equation. Therefore, with ignoring the kinetic energy, the potential energies of each links g_i (i=2,3,4,5) are given as follows:

$$g_2 = 9.849\,C_{12} + 38.411\,S_1 + 19.2056\,S_{12}$$
$$g_3 = 74.284\,C_{123} + 144.85(S_1 + S_{12}) + 99.355\,S_{123}$$
$$g_4 = 8.9817\,(S_1 + S_{12}) + 8.6362\,S_{123} + 3.1090\,S_{1234}$$
$$g_5 = 241.17\,(S_1 + S_{12}) + 231.89\,S_{123} + 166.96\,S_{1234}$$
$$+ 71.114\,S_{12345}$$

Where C_i and S_i denotes cosine and sine wave function and C_{12} denotes $\cos(q_1+q_2)$ and so on. And we abbreviate the gravity compensation of the second link (g1) so that the second link can not generate joint torques. We use the acceleration due to gravity as 9.8 [m/s2] and other known actual values which is shown in table 1 and 2 not to describe complex functions.

Finally we have the next equations of necessary joint torques t_i.

$$t_2 = 414.21C_{12} + 339.884\,C_{123} + 170.072\,C_{1234} + 71.1137\,C_{12345} - 9.849\,S_{12} - 74.284\,S_{213}$$

$$t_3 = 339.88\,C_{123} + 170.07\,C_{1234} + 71.114\,C_{12345} - 74.284\,S_{123}$$

$$t_4 = 170.072\,C_{1234} + 71.114\,C_{12345}$$

$$t_5 = 71.114\,C_{12345}$$

The suit requires only the third, fourth and fifth joint torques to lift up the patient because that the power assisting suit can actuate the air pumps at only its knee, waist and elbow joints.

For example, when the power assisting suit has the joint angles as q_i=(30,120,280,220,70) [deg], (i=1,2,3,4,5) necessary assisting force vector are represented as blue line as shown in the figure 6. The power assisting suit supports half of these weights of each links.

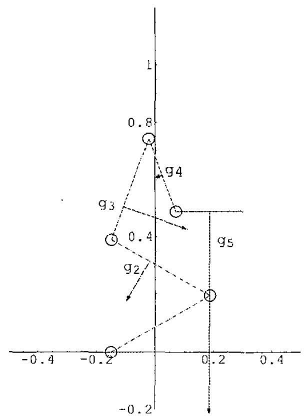


Fig.6 Normal forces of each link

5. 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.7, 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 for thin subcutaneous fat is cylinder shape and that for 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 section 8.1.

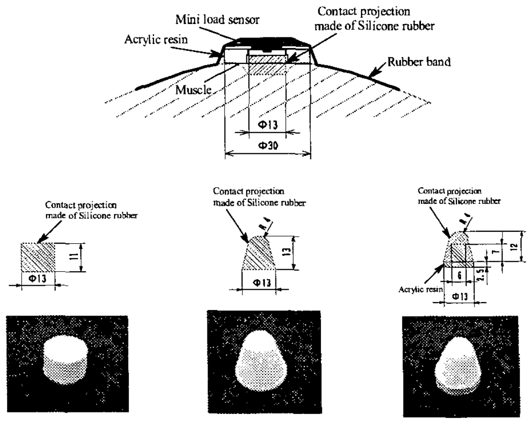


Fig.7 Muscle hardness sensors

7. Pneumatic Rotary Actuators

The pneumatic rotary actuators of elbow, knee and waist joints are shown in Fig.8. These rotary actuators are constructed of pressure cuffs (90mm×120mm) 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 [mmφ], 65 [mm] long) and the air is exhausted air by micro solenoid valves (20 [mmφ], 25 [mm] long). Two portable Ni-Cd batteries (12 [V], 30 [mmφ], 300 [mm] long) are used for this operation.

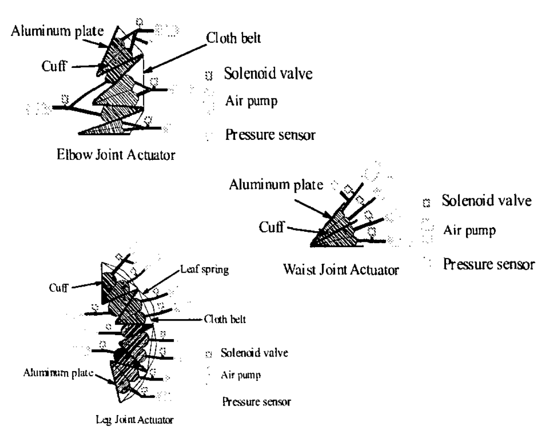


Fig.8 Pneumatic rotary actuators

8. Experimental Results

8.1 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.9. 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.

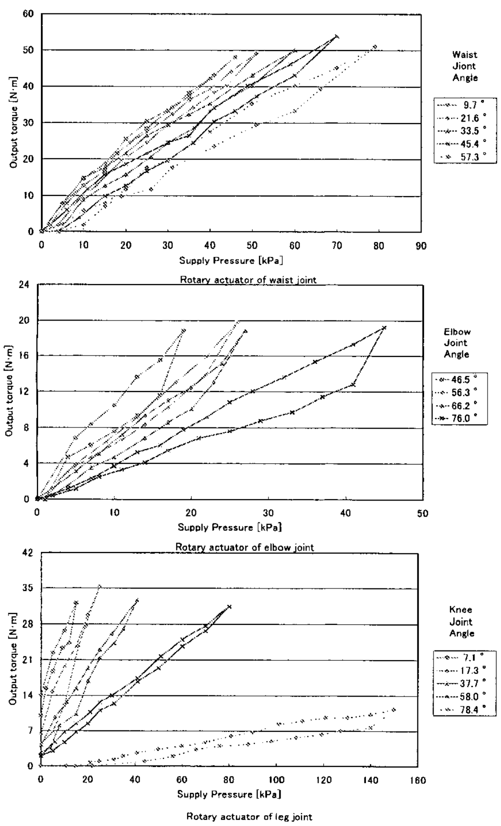


Fig.9 Characteristics of muscle hardness sensors

8.2 Rotary actuators

Fig. 10 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.

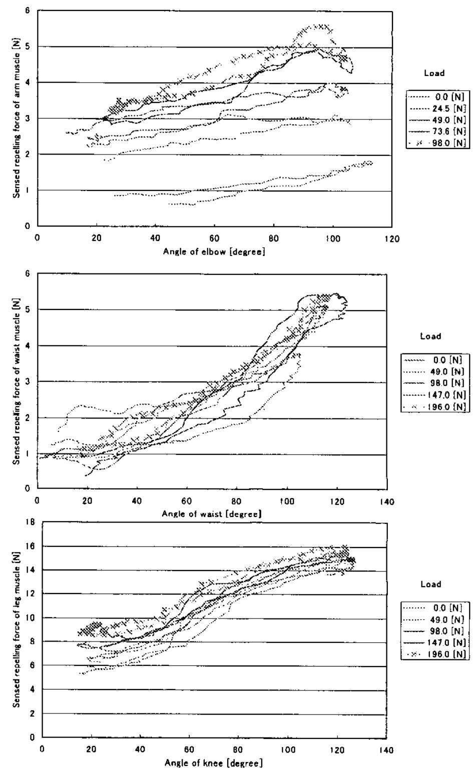


Fig.10 Characteristics of rotary actuators

8.3 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.11 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.

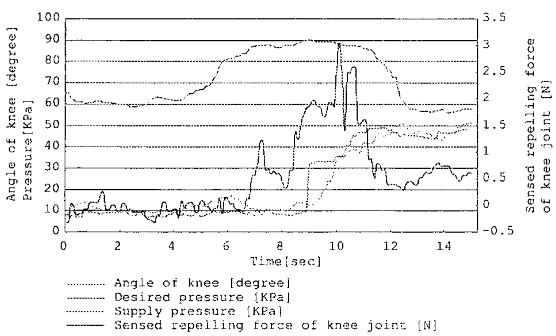


Fig.11 Operation characteristics of knee unit

Concluding Remarks

The newly stand alone wearable power assisting suit was developed and necessary joint torqueses were calculated using Lagrange's method. In addition, the muscle hardness sensor was developed and measured its characteristics. It showed the availability because of its small hysterisis and linearity. After discussing the mechanics and sensors, the suit is applied practically to a human body and a series of movement experiments that a weight of 60 [kg] on the arms are held and taken up and down was performed. According to the results of operation of lifting a person which are described in Fig.11, the gravity compensation and repelling force control works effectively.

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