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Paper:

Stand-Alone Wearable Power Assist Suit –Development and Availability–

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The stand-alone wearable power assist suit we developed gives caregivers the extra strength they need to lift patients while avoiding back injuries. To put this suit to practical use, we improved sensing system and the mechanisms. To stabilize muscle tenseness sensing more, we developed an all-in-one sensor that built the sensor into a mesh belt and improved sensing characteristics. We expanded the movable range and function of the suit. We increased actuator output torque by increasing the number of cuffs inserted into actuators. Based on equations derived from static body mechanics using joint angles, joints torques required to maintain a position are calculated by an embedded microcomputer and required joint torques was combined with muscle sensor output signals to generate control signals. We developed an exoskeleton for measurement having the same frame and potentiometers as the suit and measured muscle force by having a user wear the exoskeleton, and proved that each unit of the suit transmitted assistance torque directly to each joint. We also found that a user wearing the suit could lift weight using half or less muscle power, i.e., muscle power doubled.

Keywords: power assist suit, exoskeleton, muscle tenseness sensor, pneumatic actuator, embedded microcomputer

1. Introduction

Studies on robots for carrying people began in the 1970's, typified by Mel Kong [1] and Nurcy [2], a master-slave robot, but none were put to practical use. Support device R&D has since progressed, typified by transfer care assistance device [3] and a take-up assistance device [4] that reduced the burden on a caregiver's lower back. Both devices used a motor. Other developments include walk assistance typified by an exoskeleton leg [5] using a motor and ball screws, a raising tool [6] using air-bag pneumatic actuators, a muscle suit [7] using McKibben air actuators, and an exoskeleton leg [8] using hydraulic actuators. All were intended to assist in walking or getting up, however, not to reduce the hard labor of caregivers.

In 1991, we proposed a pneumatic power-assist suit [9] that substantially reduced the physical burden of the caregiver wearing it and developed a wearable assist suit [10–13]. We worked to eliminate of pipes and wires and used small air pumps, exhaust valves, Ni-Cd cells, and embedded microprocessors to produce a completely wearable suit [14–16].

This paper deals how we worked to make this power assist suit practical and compact while improving function. We measured assistance characteristics and confirmed the suit's practicality. By fabricating integrated detection system with sensors in a mesh belt, we developed an all-in-one muscle sensor easy to put on and take off providing stable detection. We improved in the structure and mechanism to expand the suit's movement range and increase freedom of movement. We also improved actuators to enhance assistance. We developed an exoskeleton for measurement to quantify assistance properties and conducted experiments on the exoskeleton and actual suit uses and clarified properties through comparison.

2. Power Assist Suit

The exoskeleton suit (**Fig.1**) consists of shoulders, arms, a spine, a waist, and legs. Each joint has an angle sensor (potentiometer) and the elbow, waist, and knee joints have a direct-drive pneumatic actuator having rubber cuffs covered with cloth to which small DC motor driven air pumps and solenoid exhaust valves are directly connected. This actuator generates soft assistance via air compressibility and rubber cuff elasticity that does not prevent users from moving smoothly operation. The Ni-Cd cell power source (12V, 30mm ϕ and 300mm long) is mounted on each femur of the leg.

The aluminum suit leaves a caregiver's front free to enable physical contact between the caregiver and the person being assisted.

Suit control system is shown in **Fig.2**. As the caregiver moves, muscle sensors on the upper arms, knees and back detect muscle power driving each joint. We used 50% of each joint torque calculated using a quasi-static physical dynamics model based on the angle signal of each joint, the estimated weight of the person to be assisted, and the weights of the wearer and suit as the reference value for

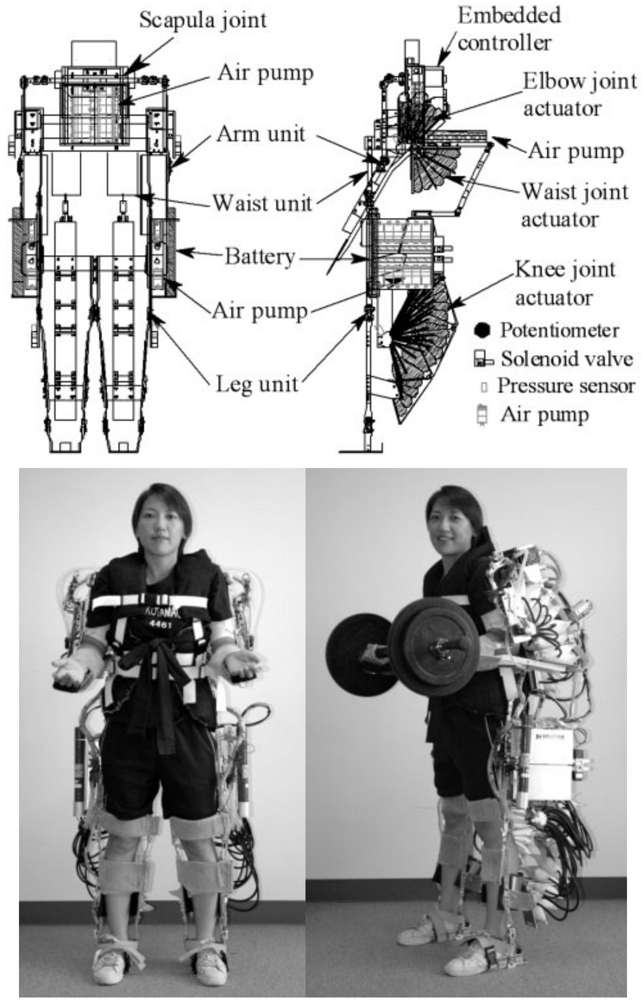


Fig. 1. Power assist suit.

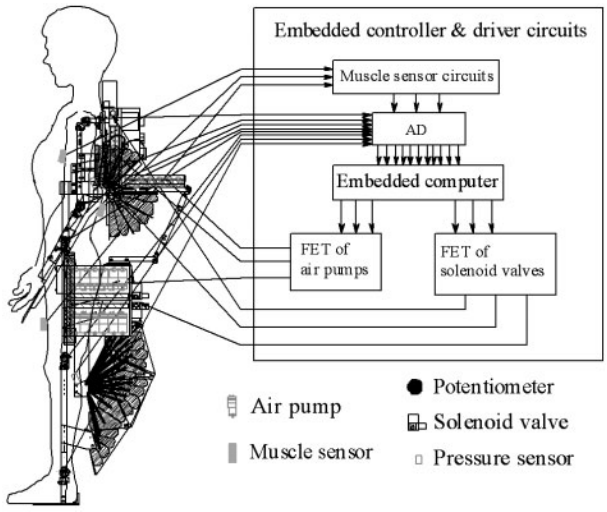


Fig. 2. Sensing and control of power assist suit.

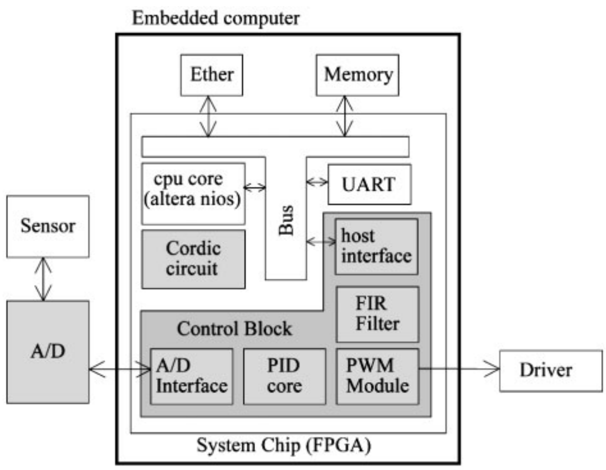


Fig. 3. Microprocessor block diagram.

assistance torque, which was added by the maximum of ±20% of the reference value for each joint torque based on the signal of the muscle sensor. This value was used as assisting torque value that should be generated by the actuator. Assuming the value calculated by the model to be T_c, the output of muscle sensor S and maximum $S_{\max}$, assistance torque is represented by the following equation:

$$T_a = T_c \cdot \left\{ 0.5 + \left(\frac{S}{S_{\max}} - 0.5 \right) \cdot 0.4 \right\}. \quad . \quad . \quad . \quad (1)$$

This assistance torque is calculated by the embedded microcomputer, and a PWM voltage signal is output to the small air pump and the exhaust solenoid valve. Air pressure in the actuator is adjusted so the actuator outputs assistance torque and auxiliary power is generated in each joint, which is needed for suit operation. The internal pressure of the actuator is PID-controlled.

3. Controller

We used a System on a Programmable Chip (SOPC) for controllers to make the suit wearable. The controller board holds a Field Programmable Gate Array (FPGA) (Altera APEX20K200E: 200K gates, $9 \times 12\text{cm}^2$ and 2.5cm high), A/D converters, the Ethernet controller, and SRAM and EEPROM.

The controller core consists of a Nios 2.0 processor (32 bits) and a control block module (**Fig.3**). The control block consists of a 24-channel PWN module (18 bits), A/D converter interfaces, finite impulse response (FIR) filters, and a PID control core (16 bits). The stand-alone block's clock frequency is 33MHz and the operation cycle is 20 clocks. The Nios processor conducts nonlinear operation, sets up the control block parameters, and provides control via a LAN.

We also developed a control block design tool to easily change the type and number of sensors and actuators based on the caregiver. The control block is represented by a combination of a 24-channel PWM (18 bits), interface logic for A/D, FIR, and PID core (16 bits). The development environment uses a graphic user interface and the user combines required functional modules on the GUI to

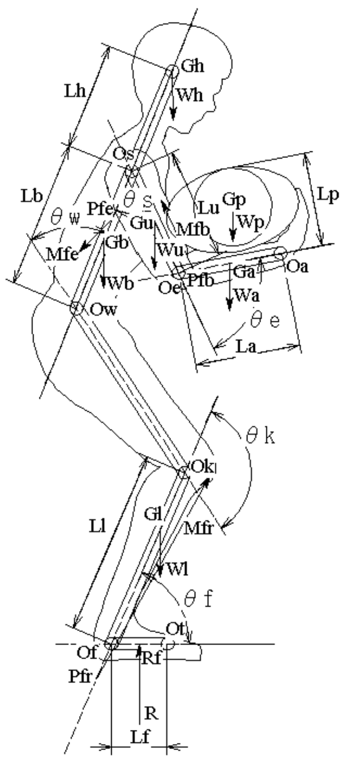


Fig. 4. Body mechanics.

build the control block.

The design tool automatically generates Verilog hardware description language (HDL) for the control block and the C program header file. The user can thus design the control block without having to be an expert. The controller for the suit designed using this tool uses approximately 85% of the FPGA.

4. Physical Dynamics Model

The role of the power assist suit is to provide torque required by the elbow, waist, and knee when a caregiver picks up or puts down the person being assisted. We developed muscle sensors to detect muscle power driving joints, but found it to be insufficient to ensure full system reliability, so we determine the physical dynamics calculation model in **Fig.4** and calculated the power of each joint drive muscle needed to maintain posture. We decided to use 50% of the torque of each joint based on the above as the reference. Since caregiver movement is slow and quasi-static, we solved the equation of equilibrium of moment with each joint angle used as a variable to find the required muscle power. Muscles generating torque for the elbows, waist and knees joints are represented by those with a muscle sensor, i.e., the arm's biceps muscle, back's erector muscle, and the thigh's rectus muscle, to determine muscle power. This model assumes that the person being assisted is supported by the center of the caregiver's arms to provide symmetrical raising and lowering.

4.1. Biceps Muscle

Muscle power M_{fb} generated by the biceps muscle is found by the equation of equilibrium of moment around center O_e of the elbow by three forces – muscle power M_{fb}, weight W_a of the lower arm and the arm unit, and weight W_p of the person being assisted.

$$M_{fb} = \left\{ \left(3W_a + \frac{7}{2}W_p\right) \cdot L_a \cdot \sin(\theta_2 + \theta_e) + \frac{7}{2}L_p \cdot W_p \cdot \cos(\theta_2 + \theta_e) \right\} \Big/ L_a \cdot \sin(\theta_e) \quad (2)$$

$$\theta_2 = \theta_s - \theta_k + \theta_f - \theta_w + 90^\circ \quad \dots\dots\dots \quad (3)$$

where θ_e is the bending angle of the elbow, θ_s the bending angle of the upper arm, L_a the distance between center O_e of the elbow and wrist position O_a, L_p the body thickness of the person being assisted, and G_a the center of gravity of the lower arm including the arm unit. It is assumed that the distance is $(1/7)L_a$ between the center O_e of the elbow and the point P_{fb} of action of the biceps muscle and that the line of action of the biceps muscle is parallel to the upper arm [17]. It is also assumed that the distance is $(3/7)L_a$ between center O_e of the elbow and the center G_a of the gravity of the lower arm including the arm unit [17], and the distance is $(1/2)L_p$ between the center G_p of the gravity of the person being assisted and the lower arm.

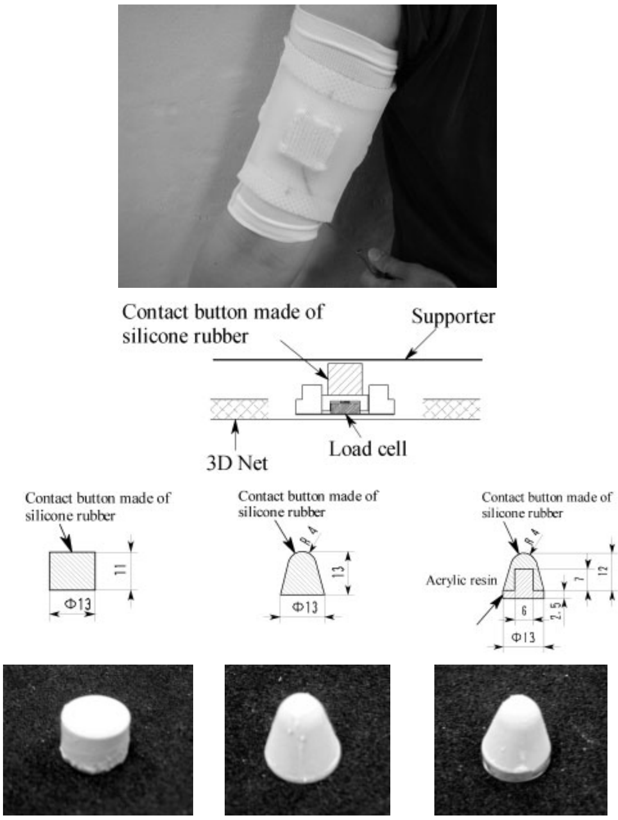
4.2. Erector Muscle

Muscle power M_{fe} that the erector muscle generates is found by the equation of equilibrium of moment around center O_w of the waist by six forces – muscle power M_{fe}, trunk weight W_b, head weight W_h, weight W_u of the upper arm including the arm unit, and weight W_a of the lower arm including the arm unit, and weight W_p of the person being assisted.

$$\begin{aligned} M_{fe} = \tfrac{3}{2L_b \cdot \sin 12^\circ} \Big[& \left(\tfrac{1}{2}W_b + W_h + W_u + W_a + W_p\right) \cdot L_b \cdot \sin\theta_1 \\ & + \tfrac{1}{2}W_h \cdot L_h \cdot \sin\theta_1 \\ & + \left(\tfrac{1}{2}W_u + W_a + W_p\right) \cdot L_u \cdot \sin(\theta_s - \theta_1) \\ & + \left(\tfrac{3}{7}W_a + \tfrac{1}{2}W_p\right) \cdot L_a \cdot \sin(\theta_e + \theta_s - \theta_1) \\ & + \tfrac{1}{2}L_p \cdot W_p \cdot \cos(\theta_e + \theta_s - \theta_1) \Big] \quad \dots\dots \quad (4) \end{aligned}$$

$$\theta_1 = \theta_w + \theta_k - \theta_f - 90^\circ \quad \dots\dots\dots \quad (5)$$

where θ_w is the bending angle of the spine, θ_k the extension angle of the knee, θ_f the extension angle of the ankle, L_b the distance between center O_w of the waist and center O_s of the shoulder, L_u the distance between center O_s of the shoulder and center O_e of the elbow, and L_h the distance between center O_s of the shoulder and center G_h of gravity of the head. It is also assumed that the distance is assumed to be $(2/3)L_b$ between center O_w of the waist and point P_{fe} of the action of the erector muscle and that the line of action of the erector muscle is 12° against the spine [17]. The distance is assumed to be $(1/2)L_b$ between center O_w of the waist and center G_b of gravity of trunk including the waist unit, and the distance $(1/2)L_u$



(a) (b) (c)

Fig. 5. Muscle tenseness sensor.

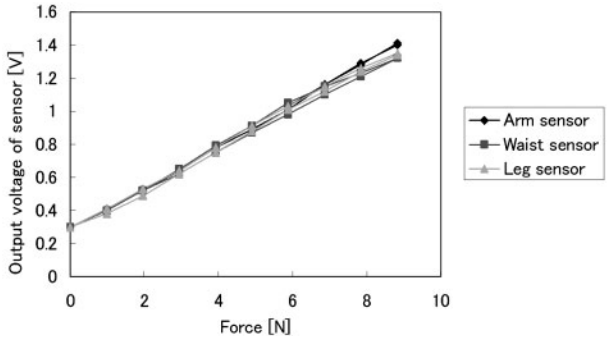


Fig. 6. Force conversion of muscle tenseness sensor.

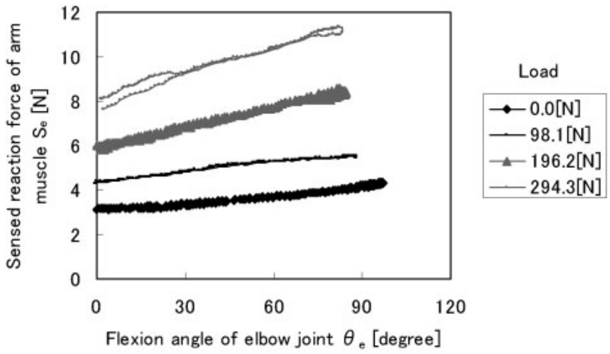


Fig. 7. Conversion of arm muscle tenseness sensor.

between center O_s of the shoulder and the center G_u of gravity of the upper arm including arm unit.

4.3. Rectus Muscle

Muscle power M_{fr} that the rectus muscle generates is found by the equation of the equilibrium of moment around center O_k of the knee by three forces – muscle power M_{fr}, weight W_l of the lower thigh including the leg unit, and floor reaction R.

$$M_{fr} = \frac{55}{58 \cdot \sin 8^\circ} \left\{ \left(R - \frac{2}{5} W_l \right) \cos\theta_f - \frac{L_f}{2L_l} \cdot R \right\} \quad (6)$$

where L_l is the distance between center O_k of the knee and center O_f of the ankle and L_f the distance between center O_f of the ankle and toe O_t. It is assumed that the distance between center O_k of the knee and the point P_{fr} of action of the rectus muscle is $(58/55)L_l$ and that the line of action of the rectus muscle is 8° against the lower thigh bone. The distance is assumed to be $(2/5)L_l$ between center O_k of the knee and center G_l of gravity of the lower thigh including the leg unit [18]. Floor reaction R_f is assumed vertical and the distance $(1/2)L_f$ between center O_f of the ankle and the line of action of floor reaction R_f.

5. Muscle Tenseness Sensor

To detect the movement of a caregiver certainly and safely, we have been developing a muscle tenseness sensor that uses muscle tenseness based on muscle power [10, 11]. We targeted ease in adding and removing the sensor and stable detection and developed an integral sensor embedded in a mesh belt (**Fig.5**). This muscle sensor is on the supporter and easy to add and remove. It is made applicable to the various thickness and subcutaneous fat of muscles at each measurement location of the caregiver by adjusting the shape, height and tenseness of a silicone rubber button pushed against the muscle.

5.1. Force Conversion of Sensor

Figure 6 shows the force conversion of each sensor for the elbow, waist, and knee using the button in **Fig.5(a)**. Each sensor had good linearity, with similar results for other buttons **(b)** and **(c)**.

5.2. Sensing of Muscle Tenseness

In experiments applying each sensor using the button **(a)** to male adults, bending and stretching of each joint was measured with loads (150 × 560mm iron plates) placed on the forearm.

Figure 7 shows detection of placing the sensor on the middle of biceps muscle, stretching the elbow ($\theta_e = 0°$) applying the load on the forearm, and bending it nearly to

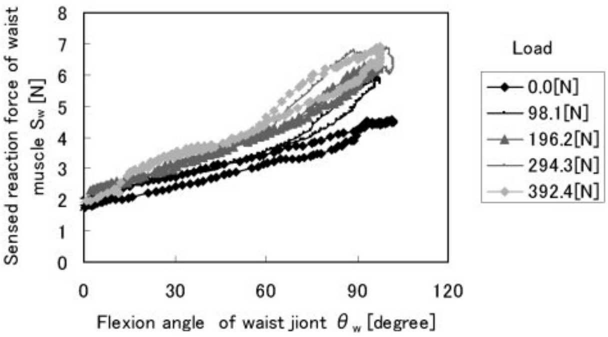


Fig. 8. Conversion of waist muscle tenseness sensor.

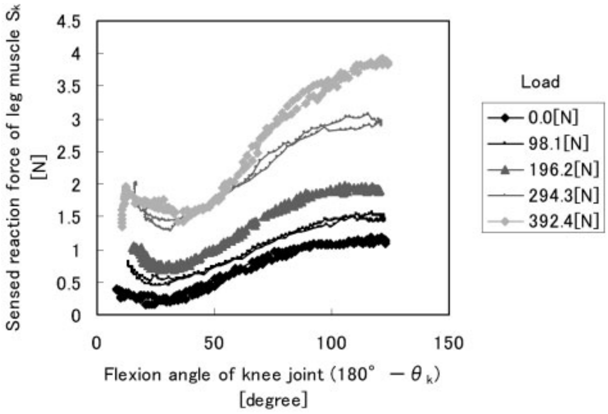


Fig. 9. Conversion of leg muscle tenseness sensor.

the right angle, then stretching it again. Proportional characteristics and load responsiveness are shown with less hysterisis and different slopes depending on the loads.

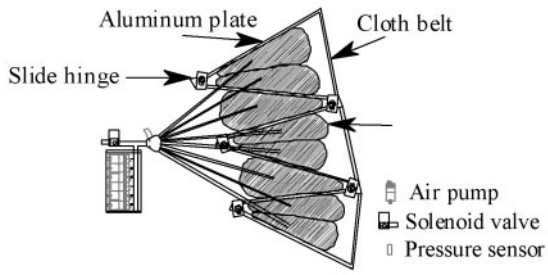
Figure 8 shows detection of placing the sensor on the lower part of the erector muscle of the waist, stretching the waist ($\theta_w = 0°$) applying the load on the forearm, and bending it, stretching it again. Proportional characteristics are shown with hysterisis and relatively fewer output changes due to the loads.

Figure 9 shows detection placing the sensor on the middle of the rectus muscle of the knee, stretching the knee ($\theta_k = 180°$) applying the load on the forearm, and bending it, stretching it again. Concave line characteristics and load responsiveness were shown with less hysterisis, different slopes depending on the loads, and the minimum where θ_k is near 155°.

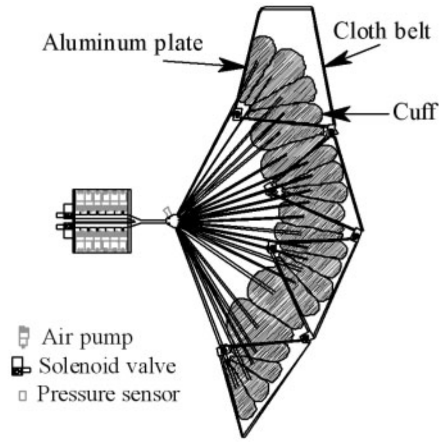
Trends among women were similar to the above, but output was lower.

6. Direct Drive Pneumatic Actuators

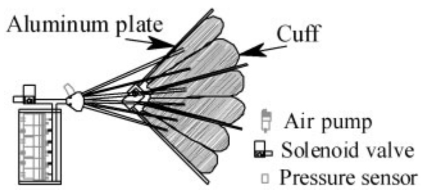
Figure 10 shows direct drive actuators developed for the elbow, knee, and waist. Those for the elbow and knee were structured so that aluminum plates were sequentially connected and folded zigzag, while the actuator for the waist used one end for connection. Commercially available cuff (90 × 120mm) for measuring blood pressure covering with cloth bag was sandwiched between aluminum plates. From connecting joints and centers of the aluminum plates with a cloth belt and interconnecting the outside with a wire [10, 11], an axis was improved to a slidable metal hinge and the periphery was connected to the cloth belt limiting the open angle. This prevented slipping of the cuff and enhanced the stability and the transfer efficiency of force.



(a) Pneumatic rotary actuator of elbow joints



(b) Pneumatic rotary actuator of knee joints



(c) Pneumatic rotary actuator of waist joints

Fig. 10. Pneumatic rotary actuator of elbow (a), knee (b), and waist (c) joints.

For the knee actuator, an alternative combination of aluminum plates 70 × 150mm and 70 × 170mm changed the shape of the actuator from a circle to an ellipse and enabled it to be slimmer than conventional [11]. Increasing the number of cuffs from 11 to 22 increased contact area between the aluminum plate and cuff, increasing output and eliminating the restoration by a plate spring needed conventionally. For the waist actuator, increasing the number of cuffs from 4 to 6 increased output. The exhaust solenoid valve remains closed during no power distribution and the internal pressure of the cuff is main-

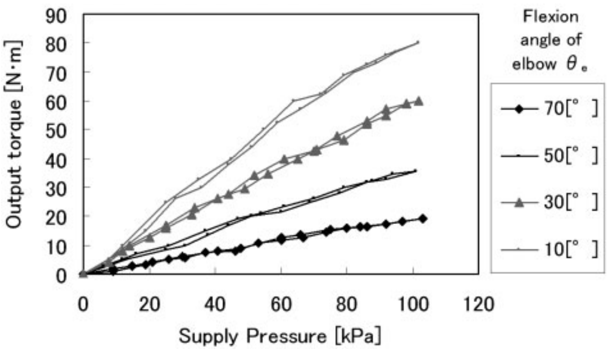


Fig. 11. Output of arm unit.

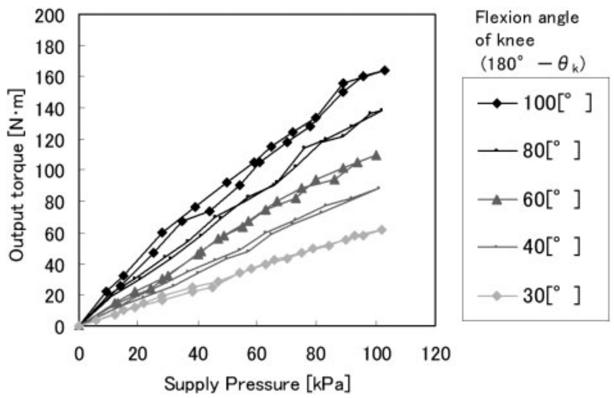


Fig. 12. Output of leg unit.

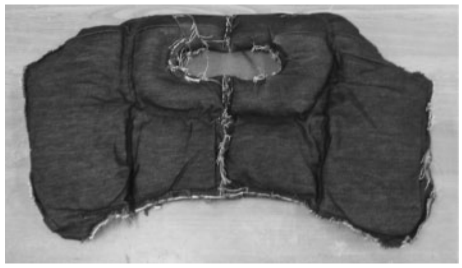


Fig. 13. Waist belt.

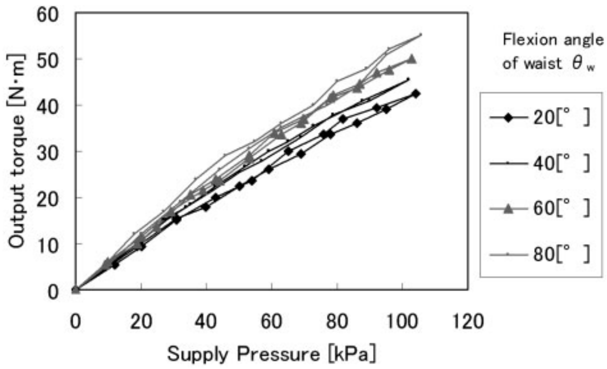


Fig. 14. Output of waist unit.

tained to ensure safety.

A small air pump (30mm ϕ, 65mm long, 6V DC, 0.1A) for air supply connected directly to an actuator has a maximum output pressure of 88kPa, a flow rate of $8.0 \times 10^{-5} m^3/s$, and a power of 1.6W ($Pa \cdot m^3/s$). A small exhaust solenoid valve (33mm wide, 25mm deep, 54mm high, 12V DC, 0.3A) has an effective sectional area of $2.5 mm^2$. A single air pump drives one to two cuffs, while a single exhaust valve empties 6 to 11 cuffs.

With the increase in the number of cuffs, the number of air pumps increased and the suit weighed 25kgf, an increase of 5kgf.

7. Units

Taking into account that the rotary center of the human joint is not a single point, joints of the auxiliary tool with the joint combining two flat gears is used for all units.

7.1. Arm Unit

As shown in **Fig.1**, an additional joint is put in for the blade bone enabling rotation of the scapulothoracic articulation, so that the hand reaches remoter place. Also, as shown in **Fig.1**, the shoulder belt for close contact between the caregiver and the unit is widened contact with the caregiver and reduced a burden on the shoulders. We developed a new belt fixed in front, so that the caregiver could put on or take off the suit. A wide belt was used to restrict the forearm.

Output characteristics of the arm unit are shown in **Fig.11**. Good linearity was provided and a maximum torque of 80N·m was obtained at a supply pressure of 100kPa.

7.2. Leg Unit

To make walking easy, we improved the sole of the foot that could be folded with the toe. We expanded the bending angle ($180° - \theta_k$) from the conventional 80° to 100°, enabling the user to crouch lower.

Output characteristics of the leg unit are shown in **Fig.12**. Good linearity was provided and a maximum torque of 163N·m was obtained at a supply pressure of 103kPa.

7.3. Waist Unit

The waist unit uses 4-node link to raise the upper body [11]. We expanded bending angle θ_w of the waist from the conventional 80° to 110°, enabling the hand to reach remoter and lower place. For the waist belt for close contact between the caregiver and the unit, we developed a belt that surrounds and protects the muscle tenseness sensor and fixed it with a wide band of "magic" tape. This made it easy to put on or take off the belt (**Fig.13**).

Output characteristics of the waist unit are shown in **Fig.14**. Good linearity was provided and a maximum torque of 55N·m was obtained at a supply pressure of 108kPa.

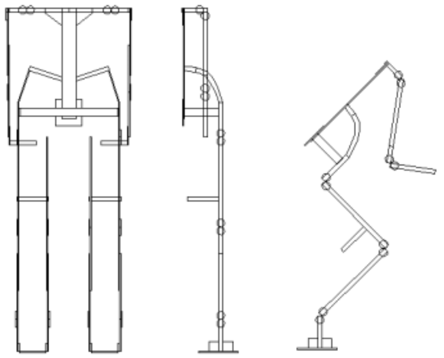


Fig. 15. Exoskeleton.

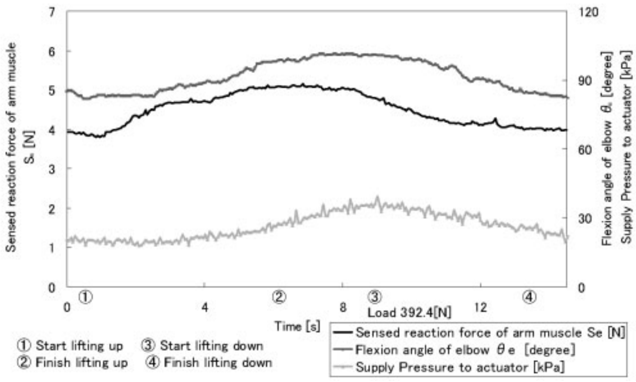


Fig. 16. Operation of arm unit.

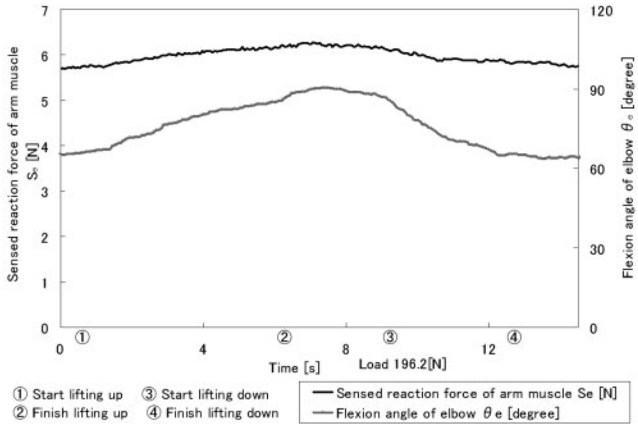


Fig. 17. Reaction of arm muscle without arm unit.

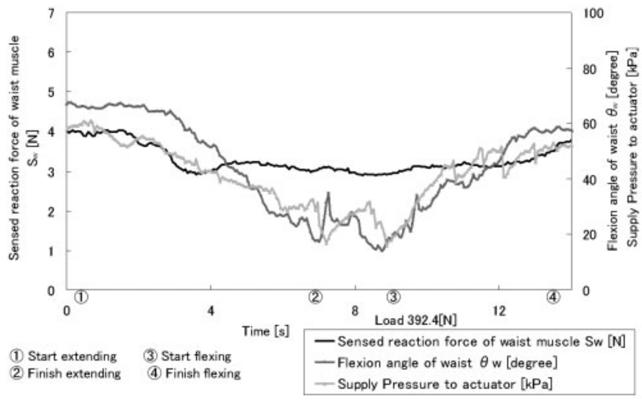


Fig. 18. Operation of waist unit.

8. Operating Characteristics of Units

Experiments confirmed that each unit was operable in the maximum movement range without giving a feeling of discomfort.

Subject put the suit on, bent the knees and waist, stretched both arms, placed 40kgf iron plate on the forearm and held it, stretched the waist while stretching the knees, raised both arms to lift it up, and then bent the waist while bending the knees, lowered both arms. Each output of the muscle sensor, potentiometer, and actuator pressure sensor was recorded during this sequential operation. Ni-Cd cells allowed a series of operations to be conducted continuously for about 20 minutes.

To examine the effect of the power assist suit, we had subject conduct the sequential operation that was the same as the operation wearing the suit, putting the exoskeleton for measurement (**Fig.15**) on, placing a 20kgf iron plate on the forearm and holding it. We recorded each output of the muscle sensor and potentiometer, and compared the data recorded wearing the suit and putting the exoskeleton on.

8.1. Arm Unit

Figure 16 shows the operation of the arm unit. Elbow angle θ_e for stretching the arm was assumed to be 0°.

The reaction of the biceps muscle increases with more energy in the arm to pick up baggage. Based on it, supply pressure to the actuator increases to assist the elbow. The muscular reaction decreases based on a decrease in the force of the arm as lowering the plate, then assistance becomes weak, and the plate is put down.

Figure 17 shows the results of exoskeleton measurement. In the reaction of the biceps muscle when the suit is worn compared to that when it is not worn, it is found to operate with a lower muscular reaction even when the weight is doubled, meaning operation can be conducted with lower muscle power.

8.2. Waist Unit

Figure 18 shows the operation of the waist unit. Waist angle θ_w for stretching the waist was assumed to be 0°.

Based on stretching the waist, the minimum is indicated at stretching posture and rises when the waist is bent again. The actuator supply pressure follows this motion and increases or decreases.

Figure 19 shows the results of exoskeleton measurement. When the waist began to be bent, sensor output immediately rose, but afterwards the same trend was seen. In the reaction of the erector muscle when the suit is worn

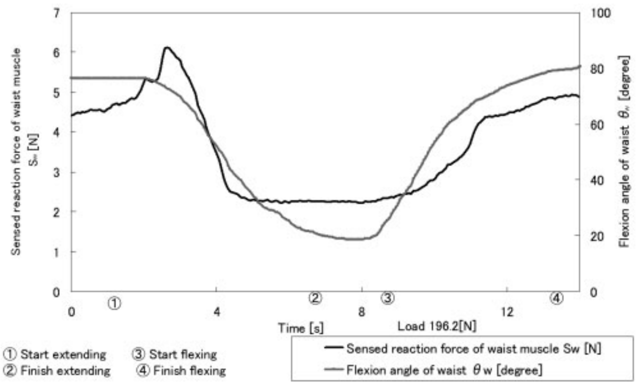


Fig. 19. Reaction of waist muscle without waist unit.

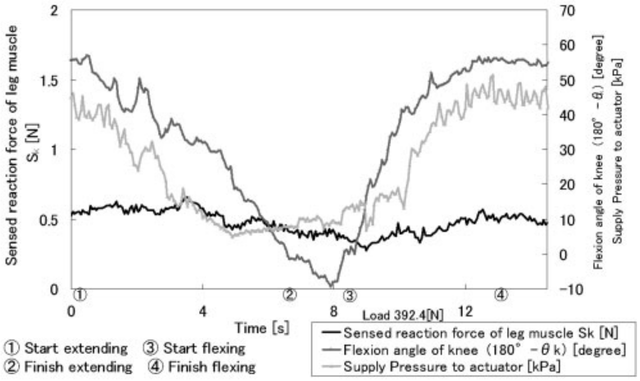


Fig. 20. Operation of leg unit.

compared with that when it is not worn, it is found to operate with a lower muscular reaction even though the weight is doubled, meaning operation can be conducted with lower muscle power.

8.3. Leg Unit

Figure 20 shows the operation of the leg unit. Knee angle (180° – θ_k) for stretching the knee was assumed to be 0°. **Fig.21** shows the results of exoskeleton measurement.

When force is applied to the leg to stand, muscle sensor output increases slightly. It decreases when standing begins and increases when crouching begins. Actuator supply pressure follows this and increases or decreases.

In the reaction of the rectus muscle when the suit is worn compared to that when it is not worn, it is found to operate with lower muscle power even though the weight is doubled, meaning operation can be conducted with lower muscle power.

9. Conclusions

We developed a muscle sensor and obtained good sensing characteristics. We improved actuators to increase their assistance and enhanced each unit to widen the movement range. We improved the suit belt to make it

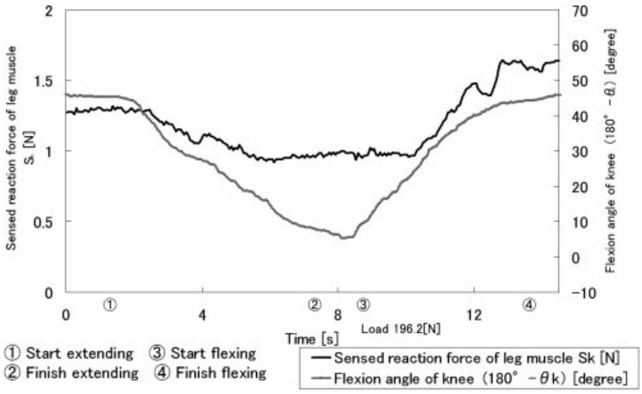


Fig. 21. Reaction of leg muscle without leg unit.

easy to put on and take off. Experiments in wearing the suit confirmed that the suit was operable in the maximum moveable range of each unit without giving a feeling of discomfort.

We developed an exoskeleton for measurement consisting of the skeleton and the potentiometer same as that of the suit to examine the effect of the power assist suit. We compared results for measuring muscle power when the suit was worn to when the exoskeleton was worn finding that each joint obtained auxiliary force from the suit to lift an item with the weight doubled but using the same or less force. This verified the effect of the power assist suit.

We are now improving the structure to make it operable in response to actual complicated care, deriving a 3D physical calculation model that corresponds to these improvements, selecting a CPU that calculates at high speed, and developing a rubber bellows actuator capable of providing softer, smoother activation.

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