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A Leg Exoskeleton Utilizing a Magnetorheological Actuator

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***Abstract* - Exoskeleton systems that can enhance people's performance or assist disabled people have been investigated in recent years. However, most exoskeletons utilize DC motors with batteries as the driving source. While the motors require a lot of power, the working time of the exoskeletons is a limiting factor for the implementation of mobile exoskeletons. This paper proposes a new leg exoskeleton that uses a magnetorheological (MR) actuator to provide controllable assistive torque. The designed MR device can function as a brake or clutch according to the working states of the user. When adjustable passive torque is preferred, the MR device will work as a brake, thus only very low power is consumed; when active torque is required, the MR device will work as a clutch, which transfers the torque generated by the motor to the user's leg. The working states of the knee joint are determined by measuring the knee joint angle and the floor reaction force. Utilizing the MR actuator, the exoskeleton will be more energy efficient.**

Index Terms* – *Leg exoskeleton, magnetorheological actuator

I. INTRODUCTION

Exoskeletons are promising in enhancing people's strength and endurance. They can also be used to help disabled people to recover their mobility, rescue workers and fire-fighters to climb buildings and run upstairs without tiring bodies, soldiers to run at higher speeds with heavy loads.

Pratt et al. developed an exoskeleton called RoboKnee [1]. It consists of an off-the-shelf knee brace and some additional structures. A linear series elastic actuator is connected between upper and lower portions of the knee brace to provide torque for the knee. This exoskeleton system could provide assistance for wearer to climb stairs and perform deep knee bends while carrying a significant load in a backpack. Low impedance was achieved by using series elastic actuators. However, this device can only work 30-60 minutes of heavy use with 4 kg of nickel-metal-hydride batteries.

The first commercialized exoskeleton, HAL-5, was designed to help elderly and disabled people walk, climb stairs, and carry things around. HAL (Hybrid Assistive Limb) is a full-body suit, which can help the user lift up to 40 kg more than they normally could. Currently, a fully charged HAL system lasts for 2 hours and 40 minutes [2]. Though it is much more efficient than the RoboKnee, but for practical applications, its working time under full charge is still insufficient.

Magnetorheological (MR) fluids comprise microscale ferromagnetic particulates dispersed in a carrier liquid. When a magnetic field is applied to this fluid, the dispersed magnetic particles will line up along the direction of the magnetic field. A shear stress is needed to disrupt this chain structure. The apparent yield stress increases as the magnetic field increases. As a smart fluid, MR fluid has several advantages such as high yield stress, good stability and fast response time. These features make the MR fluid promising for various applications. A remarkable application of MR fluids is smart prosthetics, where the MR damper is used to provide controllable damping to the knee joint and thus make the prosthesis adaptable to the person while interacting with environment. Powered by a li-ion battery, the system is able to operate for 2 days when fully charged [3]. MR damper was also used in a rehabilitation device to provide controllable resistance with excellent force tracking performance [4].

Programmable aerobic exercise equipment that uses an MR brake as controllable resistance element has been commercialized, where the MR brake can provide a maximum torque of larger than 7 Nm [5]. Zite et al. used an MR brake in orthopedic active knee brace to provide controllable resistance. However, the maximum torque produced by the MR brake is only 1.8 Nm when two magnets were used [6]. Herr and his associates developed a prosthetic knee with an MR brake. The MR brake can produce a torque ranging from 0.5 Nm to 40 Nm [7-8]. Li and his co-workers developed an MR brake used in prosthetic ankle joint to make user walk smoothly; however, the structure of the MR brake is similar to common the MR damper, and more work is needed to put this system into practical use [9].

Takesue et al. developed an MR fluid actuator, which consists of an MR fluid clutch between an input and an output parts [10-11]. This device has the benefits of safety and low inertia. However, the actuator can not provide large torque (the maximum torque is about 5 Nm), and it is a bit bulky (weight about 5 kg). Furusho and his associates developed a 3-D rehabilitation system for upper limbs with special actuators using electrorheological (ER) fluids. The ER actuator is composed of an ER clutch and its driven mechanism [12-13].

The purpose of this research is to develop a leg exoskeleton utilizing an MR fluid actuator. The new actuator will take the advantages of MR brake (to produce large torque while requiring little power) and MR clutch (to transfer torque from motor to knee with better safety). This paper is organized as follows. Section 2 describes the configuration of the exoskeleton with the MR actuator. Section 3 analyzes the MR fluid actuator. Section 4 presents the control algorithm. Section 5 gives the simulation results and discussions. Section 6 is the conclusion.

II. System Configuration

The configuration of a leg exoskeleton with an MR actuator is shown in Fig. 1, in which the controller and power supply are not shown. The main parts of the system include: braces, MR fluid actuator, sensors. The braces include upper brace and lower brace. The upper brace is bound to the upper leg and the lower brace is bound to the lower leg. The lower brace is connected to the foot, and the upper brace is connected to the waist (the upper part of the upper brace is not shown in Fig. 1). The braces transfer the assistive torque generated by the actuator to the lower limb. An elastic element is attached to the bottom of the shoe to absorb shock, thus provides protection to the system. The sensors are used to detect the user's walking condition and estimate the needed assistive torque. Two force sensors are mounted on the front and rear sole of the foot to measure the reaction force from the floor. Two strain gages are mounted on the aft and fore of the lower brace to measure the force acting on the actuator. The torque produced by the actuator can be measured using these strain gages. An angular sensor is to measure the knee joint angle. The knee angular velocity can also be derived by differentiation of the angle signal. More details about the MR actuator will be discussed in the next section.

III. MR Actuator

The structure of the MR fluid actuator is shown in Fig. 2. MR fluids are used in shear mode to produce or transfer torque. The outer cylinder is connected to the upper leg, the shaft is connected to the motor, and the motor is mounted on the lower brace. The shaft can be locked to the lower brace.

For this actuator, there are three working conditions:

1. The shaft is locked to the lower brace and the magnetic field is on. The MR device acts as a brake, which can provide controllable passive torque.
2. The magnetic field is off. The MR device doesn't work and the knee joint can rotate freely.
3. The shaft is unlocked to the lower brace and the magnetic field is on, while the motor is working. As the shaft is unlocked to the lower brace, the shaft can rotate relatively to the lower leg, thus the MR device acts as a clutch to transfer the torque from the motor to the upper leg.

The main functional components are the alternately placed outer disks (or stators) and inner disks (or rotors). The outer disks are mounted to the outer cylinder by splines. The inner disks are connected to the shaft by inner splines, which can rotate relatively to the outer disks as the shaft rotates. MR fluids are filled in the gaps between outer disks and the inner disks. As the shaft rotates and the magnetic field is on, the torque due to shear effect will be generated between the outer and inner disks. This torque is controlled by the applied magnetic field.

According to Bingham plastic model for the MR fluid, the shear stress τ can be calculated as follows

$$\tau = \begin{cases} \tau_y + \mu\dot{\gamma} & \tau > \tau_y \\ 0 & \tau \le \tau_y \end{cases} \tag{1}$$

where τ_y is yield stress, μ is viscosity of MR fluid, $\dot{\gamma}$ is the shear rate. The configuration of the disks in the actuator is shown in Fig. 3.

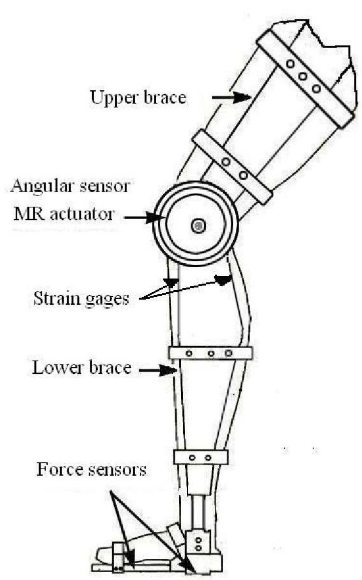


Fig. 1 Configuration of a leg exoskeleton with an MR actuator

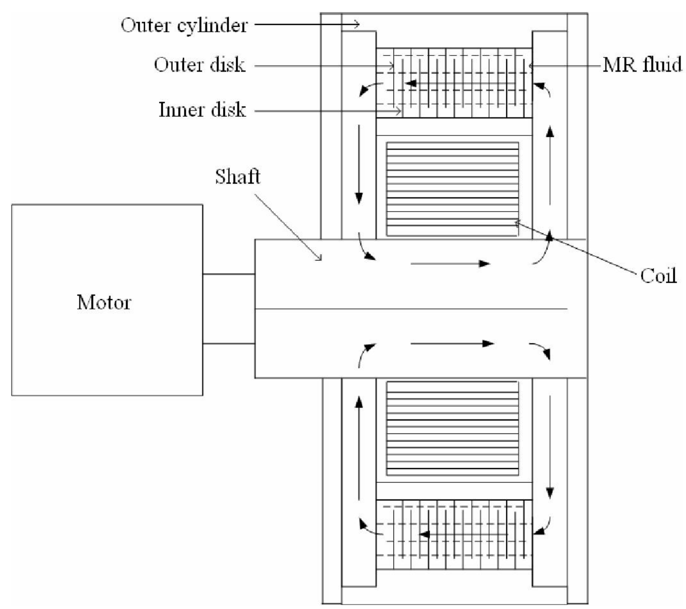


Fig. 2 Schematic diagram of actuator with MR fluid

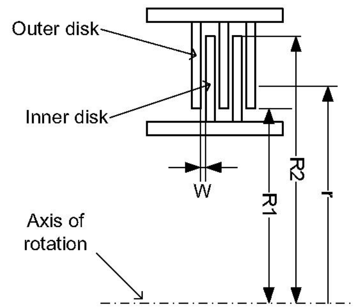


Fig. 3 The disks in the MR actuator

In this system,

$$\dot{\gamma} = \frac{\omega r}{w} \tag{2}$$

where ω is the angular velocity of the inner disk, r is the radius, w is the width of the gap between the adjacent outer disk and inner disk. Then the torque generated by the actuator under post-yield state can be derived as follows

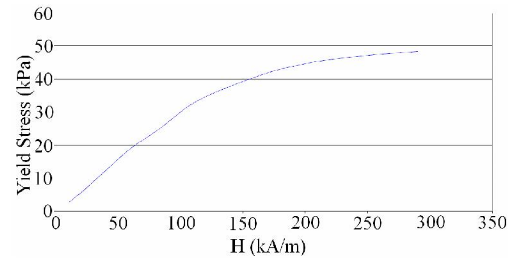
$$T = 2N\int_{R_1}^{R_2} 2\tau\pi r^2 dr = 4N\pi\left[\frac{\tau_y(R_2^3 - R_1^3)}{3} + \frac{\mu\omega(R_2^4 - R_1^4)}{4w}\right] \tag{3}$$

where N is the number of the inner disks. The first term of equation (3) is produced by the yield stress, and the second term is generated by the viscosity of the MR fluid.

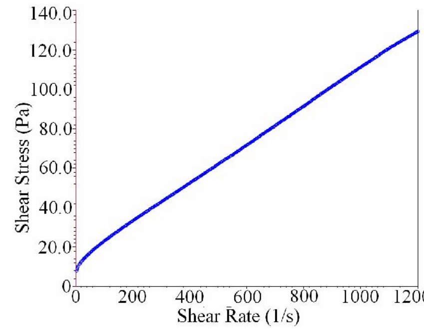
The MR fluids used here are MRF – 132DG from Lord Inc. Fig. 4 shows some of their flow properties. For the MR actuator, the parameters are chosen as follows (same as those in the patent by Deffenbaugh et al. [8]): R_2 = 0.024 m, R_1 = 0.018415 m, w = 20 μm, N = 40. The MR fluid is under post-yield condition, the viscosity μ = 0.092 Pa-s; when H = 0, τ_y = 9 Pa. For people's common movement such as walking, ω ~ $\pi/4$ rad/s, we can obtain the torque T = 0.11 Nm according to (3).

For magnetic field H = 100 kA/m, the shear stress is about 30 kPa. The torque produced by the MR device is calculated as T = 38 Nm. It should be noted that the torque caused by the viscous part is very small when the angular velocity is not large, thus it is ignored here.

Considering the friction and inertia force, the actual zero-current torque will be larger than the above calculated one. But if we reduce the mass of the inner plate and make the plate rotate smoothly (with little friction), the zero-current torque can be very small. It was proved by Herr's prosthetic knee that the MR brake has only 0.5 Nm zero-current torque [5]. When the magnetic field is on, the brake can produce a large torque. The calculated torque under magnetic field is also close to the maximum torque in the Herr's prosthetic knee, which is 40 Nm. While increasing the applied magnetic field, it is possible to obtain larger torque.



(a) Yield stress vs. magnetic field strength



(b) Shear stress vs. shear rate without magnetic field at 40ºC

Fig. 4 Properties of MRF-132DG [14]

The sensors used in this system include potentiometer angle sensor, floor reaction force sensors and stain gages. The angle sensor measures knee flexion angle (the angle is zero when the knee joint is fully extended). The angle signal is differentiated with respect to time to calculate the knee angular velocity. Knee velocity is important for determining whether the knee is flexing or extending. When knee velocity is larger than zero, the knee is flexing; while the velocity is negative, the knee is extending. The floor reaction force sensors measure the forces reacted from the floor. There are two force sensors: one mounted on the front of the foot (close to the toes) and the other mounted on the aft of the foot (close to the heel). The floor reaction force signals are used to determine whether the foot is on or off the floor. The strain gages measure the torque produced by the actuator, thus provide feedback signal for the torque control.

IV. Control Algorithm

In Herr's prosthetic knee system, the movement of knee joint during walking is divided into five phases: stance flexion, stance extension, pre-swing, swing flexion and swing extension [7]. Pre-swing is a state that starts from the knee joint fully extended during the stance state to the foot left the floor to swing. It can be viewed as stance flexion but the torque at the knee joint is small since at this state two legs support the body. For the HAL exoskeleton, the movement of knee joint during walking is divided into two phases: supporting phase and swing phase. In our system, we use three

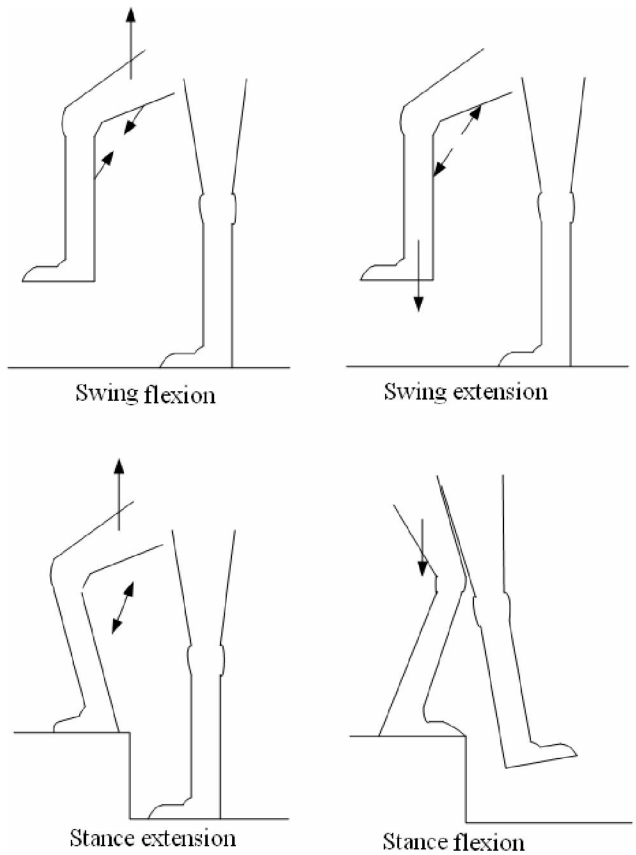


Fig. 5 Working states of knee joint

phases to descript the movement of a knee joint: swing state (includes swing flexion and swing extension); stance extension (the foot is on the floor and the knee joint extends to lift up the body) and stance flexion (the foot is on the floor and the knee joint flexes) as shown in Fig. 5.

When the knee joint is under swing state, little torque is required for the knee joint to rotate; the actuator will work in condition 2, i.e. the magnetic field is turned off and only very small torque is produced due to viscous fluid damping force and friction, etc. When the knee joint is under stance extension, the actuator will work in condition 3, i.e. the motor works and the MR device functions as a clutch to transfer the active torque to the knee joint. When the knee joint is under stance flexion, the actuator will work in condition 1, i.e. the motor doesn't work and the MR device functions as a brake to provide resistive torque.

The sensors measure the angle, angular velocity and floor reaction force to determine the state of the knee joint. The controller will shift to the state according to the movement of the knee joint. Fig. 6 shows the state transition of the knee joint. The floor reaction force (F_r) and knee angular velocity (ω_k) are compared to the thresholds F_{th} and ω_{th} to judge what state the knee joint will change to. Ideally, the thresholds are zero. However, to make the system more robust, some small values are assigned to these thresholds.

To provide appropriate assistive torque is very important. Myoelectricity (EMG) signal is used in HAL-3 as the primary command signal to provide power assist. However, only using EMG signal would cause discomfort because the time lag occurs while the estimated torque is calculated from myoelectricity. Floor reaction force is used to reduce this discomfort [15]. Pratt et al. estimated the knee torque using the following equation [1]

$$\vec{\tau}_e = \vec{R} \times \vec{F} \tag{4}$$

where $\vec{\tau}_e$ is the estimated knee torque, $\vec{R}$ is the vector from the ground reaction force to the knee joint, and $\vec{F}$ is the floor reaction force. The assistive torque is then proportional to this estimated torque.

In this system, the actuator will provide damping or assisting torque according to the estimated knee torque. The method is similar to that used by Pratt et al. The force vector is assumed to be purely vertical, and the hip is directly over the heel, thus $\vec{R}$ is determined by the knee angle and the lengths of thigh and shank. This method will lead to some error, but since the actuator only provides part of the total torque, it would not cause significant problems, and the user would quickly compensate it.

The control algorithm in this paper is a combination of state control and proportional torque control. The actuator will apply assistive torque during stance state, and the assistive torque ($\vec{\tau}_a$) is proportional to $\vec{\tau}_e$, i.e., $\vec{\tau}_a = k\,\vec{\tau}_e$. For the following discussion, we set the coefficient $k = 0.25$.

V. Results and Discussions

In this section, simulations of walking, sitting down and standing up with the leg exoskeleton are carried out. The results will be compared to the estimated torques for normal walking and standing up without leg exoskeleton. Fig. 7 shows the angle, EMG signals and estimated torque for left knee joint in normal walking and standing up.

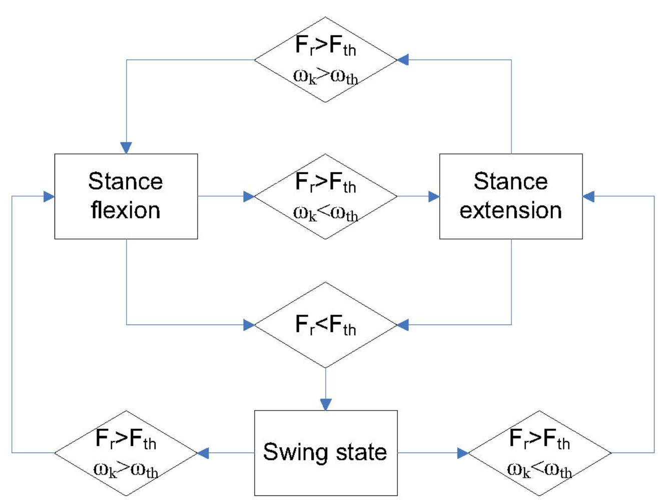
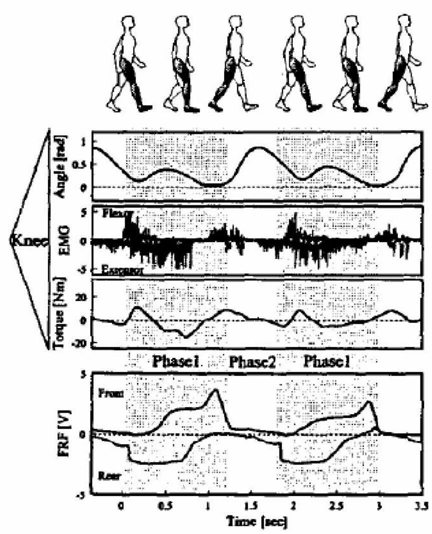
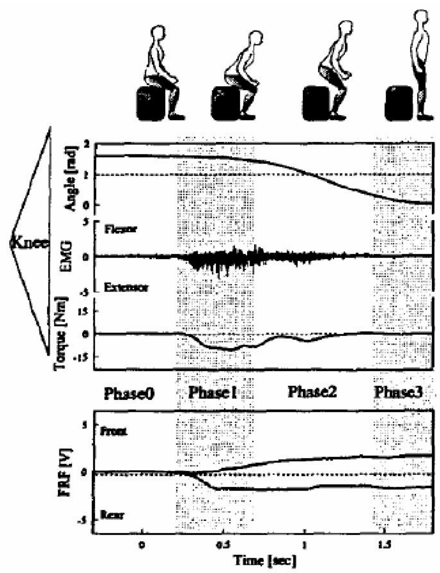


Fig. 6 State transition of knee joint



(a) Normal walking



(b) Standing up

Fig. 7 Angle, myoelectricity signals and estimated torque for left knee joint [15]

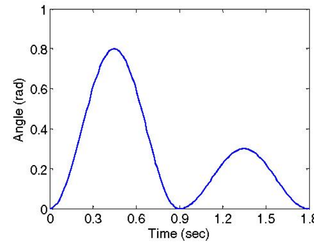
For simplicity, the variation of the knee angle α (α = 0 when the knee is fully straightened) can be viewed as two alternant sinusoidal functions.

For the swing phase,

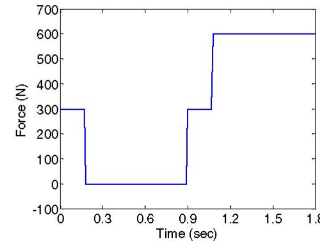
$$\alpha = 0.4\left[1+\sin(\frac{2\pi}{0.9}t-\frac{\pi}{2})\right], \; t = 0\text{~}0.9 \text{ sec} \qquad (5)$$

For the supporting phase,

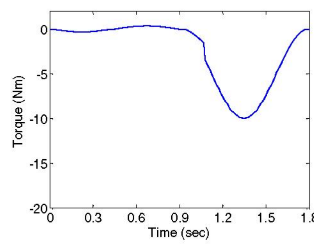
$$\alpha = 0.15\left[1+\sin(\frac{2\pi}{0.9}t-\frac{\pi}{2})\right], \; \text{t} = 0.9\text{~}1.8 \text{ sec} \qquad (6)$$



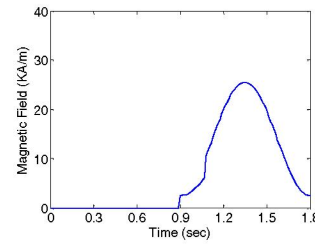
(a) Angle of knee



(b) Floor reaction force



(c) Torque under control



(d) Applied magnetic field

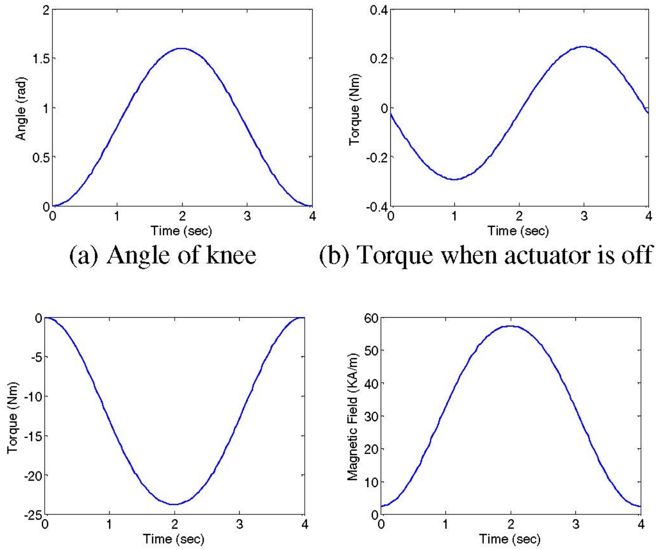
Fig.8 Angle of knee, floor reaction force, torque produced by the exoskeleton during walking

Also for simplicity, the maximum floor reaction force is regarded as the weight of the person, and the weight of the person is equally distributed to each foot during two leg supporting phase. Assuming the weight of the person is 600 N, the floor reaction force can be expressed as:

$$F_r = \begin{cases} 0 & \text{N} \quad \text{Swing phase} \\ 300 & \text{N} \quad \text{Two leg supporting phase} \\ 600 & \text{N} \quad \text{Supporting phase} \end{cases} \qquad (7)$$

Fig. 8 shows the simulation results of the user with this exoskeleton during walking. Here, the parameters are chosen as follows: length of lower leg, 0.5 m, length of upper leg, 0.4 m. The walking cycle is as follows: (1) 0 - 0.18 s: pre-swing of one leg (two leg supporting phase); (2) 0.18 - 0.45 s: swing flexion of one leg (one leg supporting phase); (3) 0.45 - 0.90 s: swing extension of one leg (one leg supporting phase); (4) 0.9 - 1.18 s: stance flexion of one leg and pre-swing of the other leg (two leg supporting phase); (5) 1.18 - 1.35 s: stance flexion of one leg (one leg supporting phase); (6) 1.35 - 1.80 s: stance extension of one leg (one leg supporting phase).

In Fig. 8 (c), from 0 to 0.9 second (the leg is in swing phase), the magnetic field is off, and the torque produced by the actuator is near zero; from 0.9 to 1.8 seconds, the magnetic field is on and the torque produced by the actuator is significant. The torque would be increased or decreased as the angle of the knee joint increases or decreases. The maximum torque during walking is about 10 Nm. It can be seen from Fig. 7 (a) that the estimated maximum torque during walking is also around 10 Nm. However, it should be noted that a coefficient of 0.25 is multiplied to the estimated torque. In other words, the estimated torque in this simulation is about four times the measured torque of normal walking.



(a) Angle of knee (b) Torque when actuator is off

(c) Torque under state control (d) Applied magnetic field

Fig. 9 Angle of knee, torque produced by the exoskeleton during sitting down (0-2 sec) and standing up (2-4 sec)

It is because that, for normal walking, part of the total torque is generated by some other joints (such as hip joint) instead of the knee joint only. Fig. 8 (d) is the magnetic field applied to the actuator during walking under state control.

Now let us consider the effect of the MR actuator on the people during sitting down and standing up. If the sitting down is regarded as the reverse of standing up (as shown in Fig. 7 (b)), the whole cycle can be viewed as a single cycle of sinusoidal function.

$$\alpha = 0.8\left[1 + \sin(\frac{\pi}{2}t - \frac{\pi}{2})\right],\ t = 0\text{~}4 \text{ sec} \tag{8}$$

When t = 0~2 second, it is the sitting down process; while t = 2~4 second, it is the standing up process. The parameters are the same as before. The whole process is two leg supporting phase, and the floor reaction force is 300 N.

From Fig. 9 (c), it can be seen that a torque is developed at the early standing up phase, and then the torque will be getting smaller. It is similar to the results shown in Fig. 7 (b). However, the amplitude of the measured torque in Fig. 7 (b) is smaller, and the curve is not smooth compared to the simulation result. It is because that, for normal standing up, the hip joint provides part of the torque. Fig. 9 (d) is the magnetic field applied to the MR device.

VI. Conclusion

The feasibility of the actuator with MR fluids in a leg exoskeleton that enhances people's performance was studied. The configuration of the exoskeleton with the MR actuator was proposed. Preliminary analysis was performed for the new system. It demonstrated that the actuator can provide sufficient assistive torque. Simulations were carried out for normal walking, sitting down and standing up processes. The results showed that the exoskeleton can provide assistance as needed. The estimation of torque has certain difference as compared to the measured torque generated by the knee joint. However, the difference can be minimized by using a proper coefficient while estimating torque. And since the human body is adaptive, it can compensate some errors. Since the MR brake can produce large torque while only consuming low power, exoskeleton with this new actuator will have longer lifetime compared to exoskeleton with the use of motors only. While the MR device functions as MR clutch, the actuator will bring advantages such as safety and good controllability.

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