Leveraging Gait Dynamics to Improve Efficiency and Performance of Powered Hip Exoskeletons

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Abstract— **This paper presents a new mechanical design for efficient exoskeleton actuation to power the sagittal plane motion in the human hip. The device uses a DC motor to drive a Scotch-Yoke mechanism and series elasticity to take advantage of the cyclic nature of human gait and to reduce power and control requirements of the exoskeleton. The mechanism creates a transmission that varies between 4:1 and infinity and does not require the motor to reverse direction when the hip joint reverses direction. This paper provides a detailed analysis of how the dynamic nature of human walking can be leveraged with this hip exoskeleton design.**

Keywords— *exoskeleton, Scotch-Yoke, gait dynamics, sagittal plane, human hip, robotics*

I. INTRODUCTION AND BACKGROUND

Exoskeletons are mechanical devices that provide power to human joints using electric, hydraulic or pneumatic actuators. They are often used to help people with limited or impaired mobility train their muscles and walk better than they would otherwise be able to. Powered exoskeletons have been used to augment and enhance the physical capabilities of humans since the 1960s when General Electric and the US military developed the Hardiman exoskeleton [1]. This exoskeleton was capable of enhancing the user's strength by a factor of 25, but it weighed 680 kg (1,500 lbs) so its applications were limited. More recently, powered exoskeletons for rehabilitation have been designed for use in medical centers and research facilities and have become much smaller and lighter than the Hardiman. Some of these exoskeletons have on-board power supplies and control units such as the Cyberdyne Hybrid Assistive Limb (HAL) [2], Argo Technologies ReWalk™ [3] and Berkeley's BLEEX [4] which was the predecessor of Ekso Bionics' eLEGS™, while other stationary clinical rehabilitation devices are tethered to a computer and do not have the power supply attached to the user (Lokomat [5] and LOPES [6][7]).

The Hardiman from the 1960s was powered primarily by hydraulics via an 18.7kW (25HP) compressor running at 3,000psi. With improvements in computation, motors and batteries, more recent exoskeletons rely primarily on electric motors. The HAL uses DC motors coupled with harmonic drive gearheads to actuate the knee joint and the hip joint in the sagittal plane. Harmonic drives provide a large reduction ratio in a small package and are used in all of the HAL actuators.

Similarly, the ReWalk™, eLegs™ and the powered orthosis presented by Farris *et al* [12] are all operated by DC motors, where the first two operate via linear actuators and the latter by a driven chain mechanism. These devices are similar to the HAL system, but with only the hips and knees being actuated. The BLEEX system powers the ankle, hip and knee joints in the sagittal plane by hydraulic actuators, using one for each joint. The stationary LOPES device uses flexible Bowden cables to actuate the hip and knee joints while the user is walking on a treadmill. The actuators and control circuitry for LOPES is located elsewhere in the room. While all of these devices actuate the hip and knee joints in the sagittal plane, they also require the actuators to reverse direction each time a joint reaches the end of its range of motion.

The contribution of this paper is the development of a continuous mechanism to actuate the sagittal plane motion of the human hip. First, the typical motion of the hip's sagittal plane during a normal gait cycle will be presented, followed by a description of the mechanism and a discussion of its benefits over traditional hip actuators. Finally, the future development of this mechanism will be discussed.

II. SAGITTAL PLANE HIP DYNAMICS

A. Torque, Angle and Power of the Hip Sagittal Plane

The cyclic nature of the sagittal plane gait cycle can be seen in Fig. 1. This data is typical of the healthy human gait cycle shown in [8]. Plots A and B of Fig. 1 show how the hip angle and torque vary in a quasi-sinusoidal manner from a peak at heel strike (0%), a low at about 50% of the cycle, and then another peak at the next heel strike (100%). The peak sagittal plane torque is 1.3 Nm/kg and the peak flexion angle is 22 degrees.

The power generated or absorbed by the hip joint is shown in Fig. 1(c). It can be seen that the hip generates positive power for the individual until 35% of the gait cycle. Positive power means the individual's muscles are generating power to propel them forward. After the 35% mark the hip angle crosses zero and corresponds to the point when the individual's leg is completely vertical. At this point, passive structures such as tendons and muscle fascia stretch and absorb energy which stops the hip from extending any further.
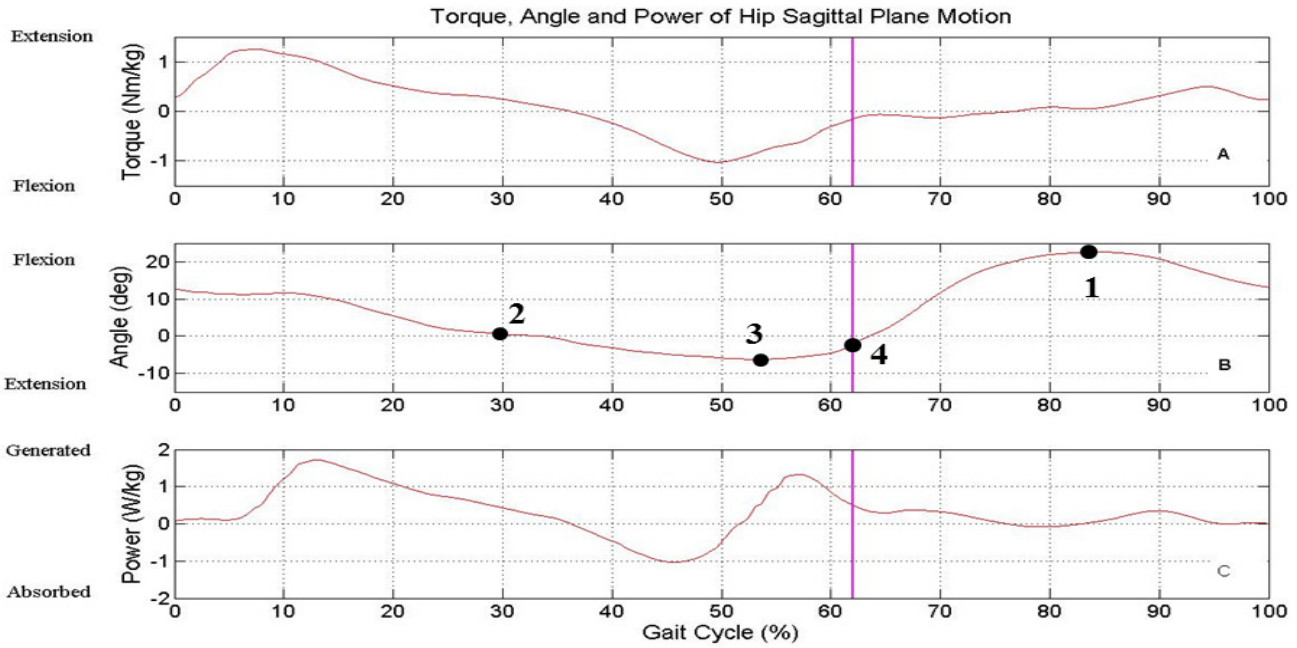
Fig. 1. Torque, angle and power of human hip joint in the sagittal plane. The solid vertical line denotes the toe-off point where the swing phase begins. The 0% and 100% points are heel strikes. Torque and power are scaled based on the individual's mass and have units of Nm/kg and W/kg, respectively. The number points in B correspond to the four phases of the mechanism in Fig. 2 Data from [8].

These structures exert the negative torque seen in Fig. 1(a). Since energy is being absorbed and dissipated by passive tissue during this phase the power generation is negative and it is possible for passive mechanical elements, such as springs, to absorb this energy instead of human tissue. Doing this would allow energy to be stored and later released during the next peak of positive power generation. This will be discussed in the next section of this paper. After toe-off (indicated by solid vertical line) the power remains minimal during the swing phase.

III.   ACTUATING MECHANISM

A. General Scotch-Yoke Mechanisms

Since the torque and angle profiles in Fig. 1 are cyclic in nature, it makes sense to use an actuator that is inherently cyclic as well. Other exoskeletons, such as the HAL, use DC motors coupled with harmonic gearheads to actuate joints. This requires reversing motor direction and overcoming significant inertia each time the joint reaches the end of its range of motion. These types of drive mechanisms also have a fixed gear ratio which means the motor must change speed dramatically when transitioning from periods of high torque and low speed demand to periods of high speed and low torque. These types of transitions happen with every gait cycle, as can be seen in Fig. 1(b) just after toe-off. The slope of this plot represents the angular velocity of the joint and it is much higher during the swing phase than at any time before or after that phase. It can also be seen that just after toe-off during the swing phase the torque requirement hovers very closely to zero. Conversely, just after heel strike Fig. 1 shows that the angular velocity is very low and the torque is at its peak. Clearly,
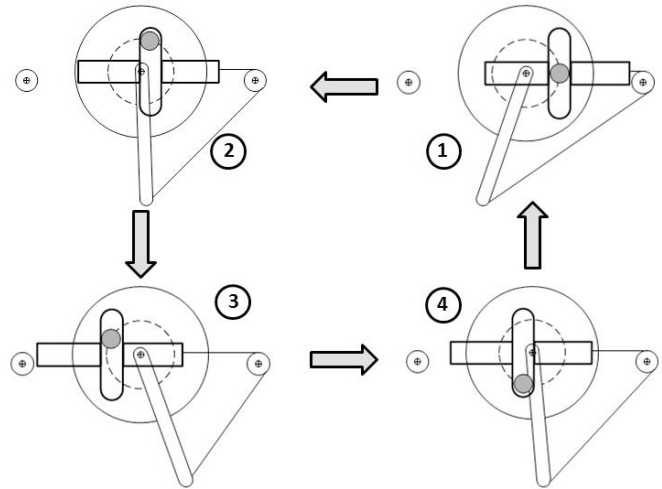

Fig. 2. Illustrates how the Scotch-Yoke mechanism moves hip from fully extended (1) to fully flexed (3). See Fig. 1 for alignment with gait cycle.

during the swing phase it would be advantageous to have a smaller transmission ratio between the motor and the joint and just after heel strike it would be better to have a larger transmission ratio. Having a variable transmission would allow the motor to spin at a more constant speed and would reduce the variation in motor current due to changes in the torque.

To realize a variable transmission that changes in a sinusoidal manner the scotch-yoke mechanism was used. Scotch-yoke mechanisms convert continuous rotary motion into reciprocating linear motion and have been used in valve control applications, internal combustion and steam engines [10-11]. These mechanisms are similar to a crank and slider mechanism

in that the linear output moves in a sinusoidal pattern except scotch-yoke mechanisms have fewer moving parts and are capable of higher torque output. Figure 2 shows a full cycle of this mechanism and it starts with the slider all the way to the right. As the wheel rotates counterclockwise the roller attached to the wheel pushes the slider to the left. After the wheel has rotated 180 degrees the slider reaches the end of its range of motion and as the wheel continues to rotate the slider begins to move toward the right. This reciprocating motion continues as long as the wheel continues to turn. When the wheel rotates at a constant velocity the slider reaches its maximum velocities when the wheel is at top dead center and bottom dead center.

B. The Mechanism

The mechanism used to actuate the hip joint in the sagittal plane is shown in Fig. 3. This mechanism takes advantage of
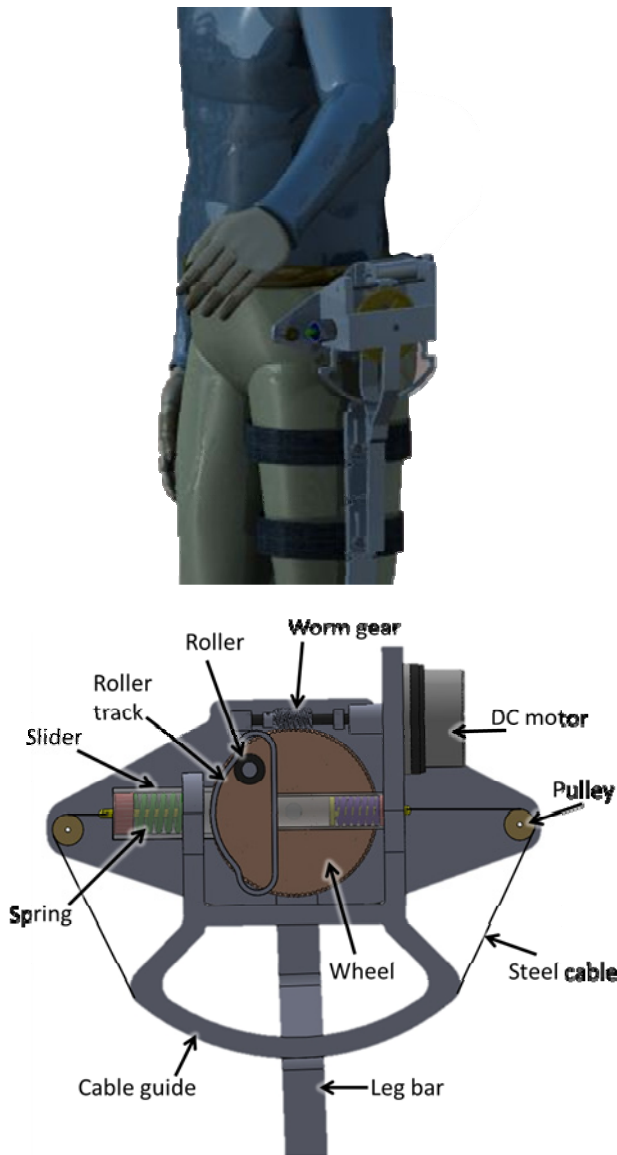

Fig. 3. Device mounted to side of user as well as detailed view of the mechanism.

the cyclically varying transmission ratio of the scotch-yoke mechanism. Periods of peak torque and low speed in the mechanism are lined up with periods of peak torque and low speed of the individual's leg. Similarly, periods of low torque and high speed of the mechanism, at top or bottom dead center, are lined up with the swing phase of the leg since this phase has the highest speed and lowest torque requirements.

A 200W brushless DC motor turns a worm gear through a 36:1 fixed reduction ratio. This worm gear turns the wheel of the scotch-yoke mechanism. Attached to the wheel is a roller that pushes on a track connected to the slider. As the wheel spins the roller pushes on the track and moves the slider back and forth. A roller is used instead of a sliding peg to reduce frictional losses and to ensure smooth motion. The slider is made up of two hollow tubes (transparent in Fig. 3 for clarity) with springs inside them to create a series elastic effect. Series elasticity is commonly used in human-machine interfaces and has many benefits such as reduced control bandwidth and improved user comfort [9]. These springs can also be used to measure the torque currently being exerted at the hip joint. Since the springs have a known stiffness and their displacement can be easily measured, the force can be determined and the moment can be computed. This eliminates the need for strain gauges to be placed along the leg bar.

Steel cables connect the springs in the slider to the vertical leg bar that attaches to the user's leg. This bar is connected to the mechanism via a pin joint located behind the wheel and as the slider moves left and right it pulls on this steel cable and causes the leg bar to move back and forth. To maintain a constant moment arm a cable guide and two pulleys located at the ends of the slider are used.

One unique feature of this actuation mechanism that doesn't appear in every scotch-yoke mechanism is the dwell that is cut into the left half of the roller track on the slider. This part of the track is cut to have the same radius of curvature as the wheel the roller is connected to, so when the roller enters this region of the track it will no longer be pushing on the track and causing it to move further to the left. This causes the slider to dwell in one location before moving again and it is in this region that the negative power generated by the hip in Fig. 1(c) can be stored for later use. Since the slider is not actuated in this region, moving the leg bar requires the spring on the left to be compressed by the motion of the user's leg. With the dwell and the spring in place the spring will be compressed as the leg goes backward and the spring will absorb the energy instead of the passive tissues in the user's hip. This energy can be released during the peak of positive power that occurs just after the negative power phase to reduce demands on the DC motor. Knowing the peak hip angle and peak torque, it is possible to compute the spring stiffness necessary to mimic the natural behavior of the user's passive hip structures and successfully store the energy they would normally absorb.

In Fig. 1(a) it can be seen that a 90kg user would have a peak negative torque of 90Nm. It can also be seen in Fig 1(b) that the hip has a maximum flexion angle of 6 degrees. With

an actuator moment arm of 15.2 cm (6 in) the linear displacement of the spring is 16.0 mm. A 15.2 cm moment arm requires 591 N of force to produce the peak torque of 90 Nm; the spring must provide 591 N when compressed 16.0 mm to provide 100% of the power needed. Since this actuation mechanism is only intended to provide 50% assistance to the user, the spring need only provide a peak of 295 N when compressed 16.0 mm. To meet this requirement, a spring with a stiffness of 18,500 N/m (106 lbs/in) is used to absorb the power dissipated during the negative power phase of walking.

C. Equations of Motion

Analysis of this mechanism is straightforward and requires a simple torque balance between the force on the wheel from the worm gear and the force on the roller from the slider. Equation (1) shows how the torque exerted on the wheel by the motor is amplified by the geometry of the mechanism as well as the current angle of rotation of the wheel.
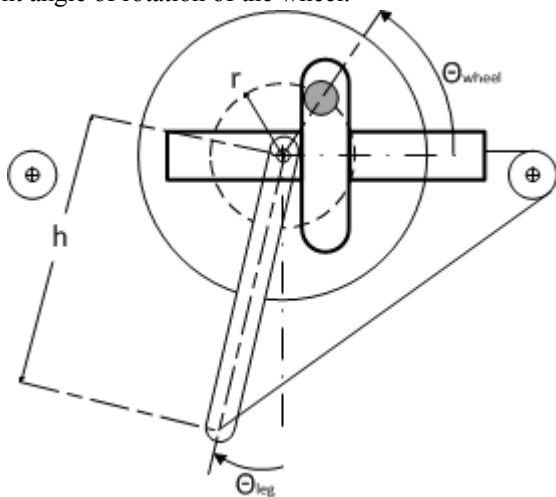

Fig. 4. Simplified sketch of the mechanism labeling important dimensions and angles.

$$\tau_{wheel} = \frac{\tau_{leg}\left(\dfrac{r}{h}\right)\sin\left(\Theta_{wheel}\right)}{\eta} \tag{1}$$

$$\frac{\tau_{leg}}{\tau_{wheel}} = \frac{h}{r\sin\left(\Theta_{wheel}\right)}\eta \tag{2}$$

where τ_{wheel} is the torque applied to the wheel from the motor, τ_{leg} is the torque required at the leg, determined from biomechanics data, θ_{wheel} is the current wheel angle, η is the efficiency, r is the distance between the center of the wheel and center of the roller, h is the distance between the center of the wheel and the point where the steel cable connects to the leg bar. Rearranging (1) yields the transmission ratio in given

in (2). It can be seen that there is a mathematical singularity in (2) for the case when the wheel angle is at either horizontal extreme: either 0 or π. This singularity creates a theoretically infinite gear ratio. In reality, there is no displacement happening at these extreme points since the slider is at the end of its range of motion, so there is no torque being exerted. Therefore, at these points the efficiency is zero. The forces at these extremes can be thought of as unstable equilibrium points, where a small displacement in either direction will require motor torque to hold it in place but exactly at the equilibrium point it is statically loaded.

The required leg torque at each point in the gait cycle is known from [8]. It can be seen from (1) that the motor must provide the smallest amount of torque when the wheel angle is at 0 or π. By making the periods of peak torque line up with these two wheel positions (7% and 50% of gait cycle from Fig. 1(a)) less demand will be placed on the motor. Additionally, (3) and (4) show how the angle and velocity, respectively, of the leg bar change with the wheel angle.

$$\Theta_{leg} = \left(\frac{r}{h}\right)\cos\left(\Theta_{wheel}\right)+\Theta_{leg\_offset} \tag{3}$$

$$\dot{\Theta}_{leg} = -\left(\frac{r}{h}\right)\sin\left(\Theta_{wheel}\right)\dot{\Theta}_{wheel} \tag{4}$$

where Θ_{leg_offset} is the leg angle when the wheel angle is zero. This is adjusted when mounting the leg bar to the mechanism to ensure peak transmission ratio of the mechanism lines up with the peak torque regions of the gait cycle.

It can be seen from (4) that the leg reaches its maximum velocity when the wheel angle is at $\pi/2$ or $3\pi/2$; 90 degrees out of phase with the angles of peak torque. This means that during the wheel's rotation, the regions of peak torque and minimum velocity are offset from the regions of minimum torque and maximum velocity, just like it is in the human gait cycle in Fig. 1.

It can be seen in (3) that the maximum and minimum leg angles are controlled by the ratio of r/h and by the leg offset angle. Both of these parameters can be adapted for the specific user when the device is being fitted to them. Changing where along the leg bar the cable connects will change h and therefore change the ratio, and changing the length of the cable will change the leg offset angle. These adjustments allow the device to work with a wide range of stride lengths and gait patterns.

IV. PERFORMANCE

This section will analyze and discuss how the mechanism described in the preceding section reduces demands on the motor as compared to more standard actuation mechanisms. The reader should note that the mechanism is designed to leverage the dynamic patterns of walking. While it is possible to oscillate the motor direction and not complete a full rotation to actuate small steps, such as those taken when walking first begins or when stopping, efficiency will be lost and the control of the device becomes more complicated. Since most of a
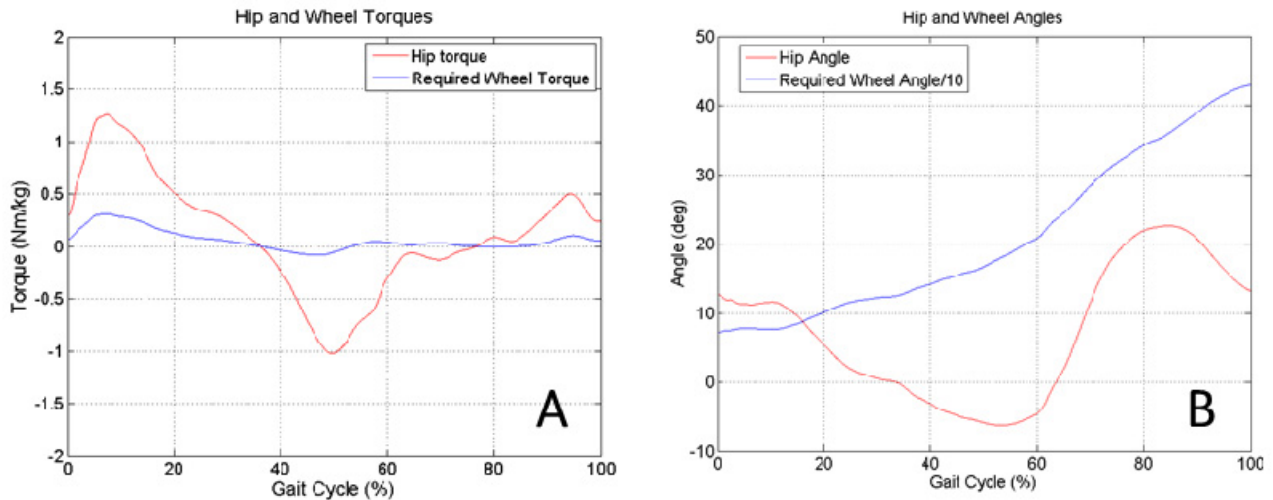
Fig. 5. A) Leg torque at hip joint and torque required on wheel of mechanism. Torque curve is significantly reduced and flattened over the gait cycle by using scotch-yoke mechanism.   B) Shows how leg angle at the hip in the sagittal plane changes over the gait cycle. Hip angle changes from positive to negative each time the leg is vertical but the mechanism's wheel angle always has the same sign. This means the mechanism does not have to reverse direction to move the leg back and forth. Note that the start and end angles of the wheel are equal because heel strike happens at both 0% and 100% of the gait cycle. The wheel starts at an angle of 72 degrees and ends at an angle of 432 degrees, but by subtracting 360 degrees from 432 it can be seen that the start and stop angles are the same.

person's walking time is a spent dynamically walking and not taking an initial step or stopping, this is an acceptable trade off.

A. Reduction in Torque Variation

Figure 5(a) compares the required leg torque to the required wheel torque over the gait cycle. It can be seen that despite large variations in the required leg torque, the torque required on the wheel from the motor does not change much. It has a peak of 0.31 Nm/kg just after heel strike and does not go above 0.1 Nm/kg at any other point.

Having a nearly flat torque curve over the entire gait cycle reduces the demand on the motor and worm gearing system. A more uniform force improves the fatigue strength of the gear teeth and improves the life of the motor. It also reduces the demands on the electrical system and reduces the need to design a system capable of handling frequent large current spikes. Furthermore, the drive motor can be slightly undersized since it is only required to provide a large amount of torque for roughly 10% of the gait cycle. This allows the motor to be pushed up to its power limits because it will have the other 90% of the cycle to cool down again.

B. Reduction in Speed Variation

Along the same lines as the previous section, the scotch yoke mechanism reduces the variation in the velocity of the wheel as well. Figure 5(b) shows that while the slope of the leg angle curve (velocity) changes quite significantly over the gait cycle, the slope of the wheel's angle curve steadily increases.

Another significant benefit of this mechanism is that the motor only spins in one direction. In Fig. 6 it can be seen that

while the leg angles changes sign, the required wheel angle is always increasing and thus always spinning the same direction. This greatly reduces the demands of the control system because the motor inertia does not need to be overcome every time the leg reaches the end of its range of motion; this mechanism creates reciprocating linear motion from continuous rotary motion.

V.   CONCLUSION AND FUTURE WORK

This paper has presented a scotch-yoke mechanism for continuously actuating the human hip joint in the sagittal plane.

The mechanism was introduced, explained and its performance was evaluated by comparing the torque and velocity requirements of the hip joint and actuator. It was seen in Fig. 5(a) that the highly variable torque profile of the hip joint can be reduced and flattened by using this type of mechanism.

The next step is to adapt the mechanism to provide assistance to help the user sit and stand. The current mechanism only provides assistance during normal walking but adaptations for sit-to-stand and stand-to-sit transitions exist. One option for doing this would be to passively stretch one of the two springs in the mechanism as the user sits. The potential energy lost by the user lowering themselves would be stored in one of the springs and later released when they wanted to stand back up. Future work will continue with the development and testing of a physical prototype.

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