Design of the Robotic Tendon

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***Abstract*— A Robotic Tendon is a spring based linear actuator in which the properties of the spring are crucial to its successful use in a gait assistance device. Like its human analog, the Robotic Tendon uses its inherent elastic nature to both reduce peak power and energy requirements for its motor. In the ideal example, peak power required of the motor for ankle gait is reduced from 250 W to just 77 W. In addition, ideal energy requirements are reduced from nearly 36 Joules to just 21 Joules. Using this approach, the initial prototype is expected to provide 100% of the power and energy neccessary for ankle gait in a compact 0.95 kg package. This weight is 7 times less than that predicted for an equivalent direct drive approach.**

I. INTRODUCTION

In the United States one in five persons live with some form of disability and 61% of these suffer from either a sensory or physical disability[1]. As a person ages, so increases their chances for becoming disabled. Within our growing elderly population, 20 to 50% are afffected by abnormal gait, i.e. walking impairment[2]. Abnormal gait in the elderly does not have a specific cause, however many age-related factors can affect normal locomotion. Some examples include; 1) muscle weakness, 2) slow reaction times, and 3) impaired tactile sensation from the feet.

The ability to balance is the first requirement for successful gait. The factors that affect gait also affect a person's ability to balance upright. Impaired sensory information, long processing times and weak actuation all lead eventually to an unstable balance control system. For the complex tasks of balance and gait, significant deficiency in any of these factors pushes the limits of postural stability to 'marginal' at best. The result of these factors is an increase in duration of double-limb support during gait, which leads to a decrease in walking speed. This decrease is approximately 20% less than the speed of a typical young adult[3]. Interesting to note is that this decrease in speed is not due to a reduction in cadence (i.e. frequency of gait), but is attributed to a decrease in stride length, or reach. The term 'cautious gait', coined by Nutt et. al.[4], describes this phenomenon as the response to a "real or percieved disequilibrium". Cautious gait is the result of apprehension to falling.

Many disabled individuals could benefit from some form of robotic intervention. A robotic device could provide strength where there is weakness, respond to stimuli quickly rather than slowly, and a wearable robot could sense problems early, rather than after it is too late. A robotic device in assistance to the disabled elderly can provide a solution to problems and ailments of getting old.

II. BACKGROUND

The most effective form of robotic intervention would be a wearable system that could provide the strength and performance augmentation to a person in need. However, use of the term 'wearable' implies that such a robot be portable, lightweight and most importantly safe. In order for such a device to be accessible for home use, the additional implications are that the wearable robot be economical and easy to operate. In contrast, a factory floor robot is none of these things, so simple adaptation of existing technology is not possible. In order to handle the needs of the disabled population, actuated wearable robots that are portable, lightweight, safe, economical and simple to operate are required[5].

The prevalence of powered assistance devices for the weak and elderly can be seen almost every day. Powered-seated scooters are increasingly popular and are available from a variety of commercial sources. Often these scooters even require additional modifications to ones home and automobile to accommodate their use. The popularity of the seated scooter is testament to the need for powered assistance; however, the use of these devices are in direct conflict to the belief that long term health is maintained by the inclusion of "the types of activities that provide an adequate load-bearing stimulus"[6]. Powered assistance is required, but should come in a form that promotes and supports standing/walking activities. To maintain general health and wellbeing load-bearing walking is essential.

However, the solution to developing a walking assistance robot is not trivial. It is well known that such a system would need the ability to produce large torques and be capable of high power. Such requirements raise the threshold for wearable robots to be successful in this application. Nevertheless, work in this area has already begun.

Well known projects in the area of assisted locomotion are the BLEEX (Berkely Lower Extremity Exoskeleton)[7] robot and the HAL-3 (Hybrid Assistive Leg)[8][9] robot. Both devices are rigidly attached to the wearer and are directly driven, i.e. no complaint interface. The BLEEX robot uses hybrid hydraulic actuators to drive the system, whereas the HAL-3 robot uses DC motors and gearboxes to provide power for movement to the user. The goals of the individual projects are different, but conceptually they each provide the same solution, to provide directly, both positive and negative forces to the user to achieve a desired movement pattern. For example in gait, sometimes the robot needs to push the user (positive) and sometimes for support the robot needs to resist the user (negative) and in either case the robot is putting power into the system.

In other work, a robotic powered knee, RoboKnee[10], and an active ankle foot orthosis, AAFO[11], have been developed to assist with an individual's gait. Each of these devices feature the linear Series Elastic Actuator[12] as the means of robotic control. The linear series elastic actuator features a helical spring in series with a ball screw mechanism, similar to the actuator developed by Sugar and Kumar[13] for grasping tasks. For the series elastic device, the inclusion of the spring aids greatly in force and impedance control task stability. However, even though the device uses a spring between the actuator and the environment(i.e. human), the compliance of this system is derived mostly from its controller. Based upon the geometry and length of the springs used, very little deflection or compliance would be possible in a passive situation and thus is still very nearly a directly driven system.

It has been well known to the legged robot community that the inclusion of springs in robotics can effectively reduce both the power and energy requirements demanded of an actuator[14][15]. This is because a spring can store and release energy efficiently during cyclic repetitive tasks and the power released from a spring is limited only by the natural frequency and stiffness of the system[1]. In other literature, van den Bogert describes a theoretical and passive mechanism that reduces peak power for human gait by more than 70%[16]. The passive device uses a series of elastic cords and pulleys around multiple anatomical joints to accomplish reduced power requirements. As written, the specific implementation described would not likely be practical, but the point of including springs in the design of wearable robotic systems is beneficial.

In order to meet the demanding requirements stated above, a wearable robotic device must include lightweight, energy conservative, power reducing springs to be both portable and inherently safe.

III. HUMAN GAIT

If the goal of a wearable robot is to aid a person in locomotion, then it is necessary to understand normal human gait. Gait is the term used to describe the locomotion of legged animals. Gait is a reoccurring pattern of leg and foot movements, rotations, and torques. Due to its repetitive nature, the discussion of gait is done in terms of percentages of a gait cycle. A gait cycle is defined for a single leg and begins with the initial contact of the foot with the ground or 'heel strike', the conclusion of a cycle occurs as the same foot makes a second 'heel strike'. The end of one gait cycle is of course the beginning of another. To illustrate a typical pattern of gait, consider the kinematics and kinetics of a normal ankle[17], figure 1. Notice that the ankle moment (torque) data is normalized by body weight, kg.

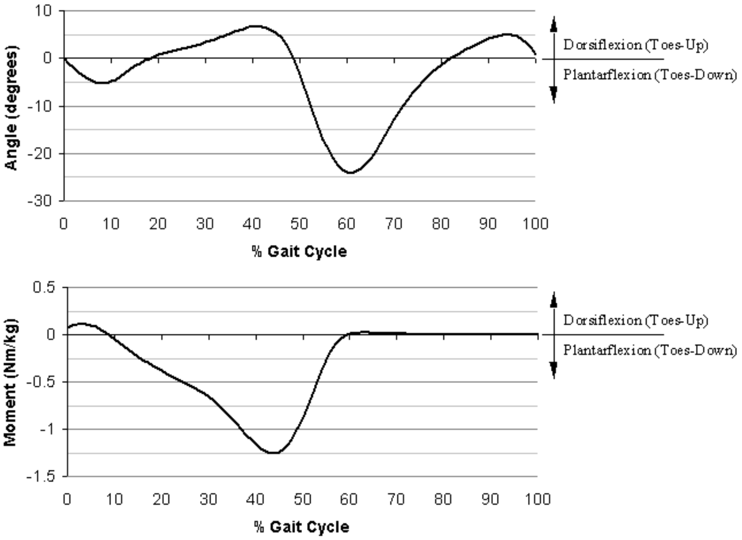

Fig. 1. Normal Ankle Gait: Kinematics and Kinetics.

In this figure, peak ankle moment occurs at roughly 45% of the gait cycle and at a value of -1.25 Nm/kg or for a 80 kg person, -100 Nm. The negative sign represents the physiological direction for which the moment occurs, in this case peak moment is acting to move the foot in a toes down direction. As an interesting note, at the point at which the peak moment occurs, the ankle angle begins a rapid desent to its lowest overall value of -24^o at 60% of the gait cycle. The region of gait approximately between 45% and 60% of the gait cycle is known as 'push off'. At the conclusion of 'push off', now considered 'toe off', the leg initiates 'swing' and the foot is then positioned for the next 'heel strike'.

IV. ROBOTIC TENDON

Use of the term Robotic Tendon implies an analogy to human physiology. Mentioned earlier, the simple inclusion of a spring to a linear actuator can provide energy and power savings to the design of a wearable robotic device. The premise of the following development is that the human muscular system uses the advantages inherent in its elastic nature. Those advantages are a minimization of both work and peak power. In terms of an electric motor, minimizing peak power implies the reduction of requirements for motor size and thus weight. Minimizing work implies a reduction of stored energy supply necessary to fulfill the demands of gaits. For a portable robotic system, these are both very important considerations.

The idealized structure of the Robotic Tendon is similar to that of the devices by Sugar and Kumar[13] and Robinson, et. al.[12]. Each of these devices include a linear actuator in series with a spring. Different from the previous devices are that estimations of the environmental forces and displacements

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[1] The released energy from a spring can reach its peak power in a period of time inversely proportional to the natural frequency of the system, $t = \frac{\pi}{2\omega_n}$, where $\omega_n = \sqrt{\frac{K}{m}}$, thus the peak power released from a spring is $P_{max} = \frac{1}{2}Kd^2\omega_n$ (K is spring stiffness, m is system mass & d is the spring deflection).

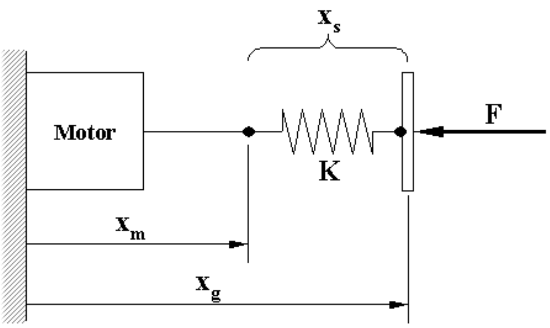

Fig. 2. Robotic Tendon Model: motor and spring in series.

are known and in our example are based upon normal ankle gait patterns. A conceptual model of the Robotic Tendon can be seen in figure 2. For this initial model, the environmental perturbations are known (i.e. gait) and the inertial effects of the linear actuator are assumed negligible.

In contrast to the direct drive example mentioned previously, a spring base actuator design will have very different characteristics. Using the simple model of the Robotic Tendon in figure 2, comparisons to direct drive approaches can be seen. In a direct drive approach, the stiffness K can be considered nearly infinite, thus all of the environmental displacements must come from the linear actuator. In the Robotic Tendon model, the selection of K is key to the energy and peak power savings in the designed linear actuator or motor requirements.

From figure 2, a development of motor power requirements based upon stiffness K can be derived. The position of the environment, x_g, is given by converting the joint angles of gait to linear displacement using a simple lever arm. The position is thus a combination of the position of the motor, x_m, and the position of the spring, x_s, see equation 1.

$$x_g = x_m + x_s \tag{1}$$

However, since a spring is a passive device its position is determined by the force, F, applied to it. Consider the basic Hookean spring shown in equation 2.

$$F = K \cdot \Delta x_s, \tag{2}$$
$$\text{where, } \Delta x_s = d_o - x_s$$

The free undeformed length of the spring is represented by d_o and is simply an offset value. Solving equation 2 for x_s, yields:

$$x_s = d_o - \frac{F}{K} \tag{3}$$

The length of the spring is based upon the environmental force and spring stiffness. Now equation 3 can be substituted into the equation for environmental position, x_g, and solved for motor required position, x_m. From this substitution the following relationship is determined, see equation 4.

$$x_m = x_g + \frac{F}{K} - d_o \tag{4}$$

and taking its derivative, yields velocity required.

$$\dot{x}_m = \dot{x}_g + \frac{\dot{F}}{K} \tag{5}$$

Knowing the forces, F, required by the gait cycle and knowing the motor's required velocity, $\dot{x}_m$, the relationship for motor power, P_m, can be obtained. Power is simply force multiplied by velocity, thus multiplying F by equation 5 will yield a relationship for motor power.

$$P_m = | \underbrace{F \cdot \dot{x}_g}_{\text{gait power}} + \underbrace{\frac{F \cdot \dot{F}}{K}}_{\text{spring power}} | \tag{6}$$

Human ankle gait power can be both negative and positive. When it is negative, a resistance motion is applied to the ankle. When it is positive a propelling motion is applied and all is a natural part of human gait. A motor unit cannot typically provide negative power, therefore it must provide power to both resist and propel human motion. For this reason, an absolute value in equation 6 is used. In addition, values for force, F, velocity, $\dot{x}_g$ and $\dot{F}$ can all be determined from human gait analysis data, thus stiffness, K, becomes the only design parameter.

To design an assistive robotic device for gait, understanding motor velocity and power requirements is fundamental. Consider the case where spring stiffness, K, is nearly infinite (i.e. direct drive). In this example the spring power term drops to zero and the motor must provide the absolute value of normal gait power. In the opposite case, consider a spring with a stiffness near zero. It is obvious to see that in the second example that the power requirements tend toward infinity. If we were to assume a straight line between these two cases it would appear that one could never do any better than a direct drive scenario. Fortunately, this simplistic relationship is not the case. On the contrary, if a spring is properly selected both energy and peak power required to perform human gait can be drastically reduced compared to the direct drive analogy.

A. Direct Drive Scenario

In order to evaluate equation 6, some assumptions about the human operator and device must be made. For this analysis, consider a 80 kg person, who has a walking rate of 0.8 Hz. Also consider that the lever arm necessary to convert the rotational ankle joint characteristics to linear movements is 12 cm. For the range of rotational displacements in ankle gait (less than 35^o), a linear movement approximation is acceptable. With these assumptions peak power for human gait is nearly 250 W.

Looking back at the angle and moment relationships for human ankle gait, figure 1, it can be seen that the highest moments and highest velocities must occur at approximately

50% of the gait cycle. As stated before, with our assumptions peak power for ankle gait is almost 250 W. To match that peak, a motor of significant size and weight would be required. As an example, the Maxon motor RE75 (Maxon Precision Motors, Inc., San Diego, CA) is rated for 250 W continuous power (rated peak power, 393 W) and weighs 2.8 Kg not including the weight of a gearbox. Adding an appropriate gearbox would increase the weight by 3.8 Kg; this combined weight is no small consideration for a portable wearable device. Although the peak power requirement for gait is high, it is only at this magnatude for the instant at which 'push off' is initiated. For the remainder of the gait cycle the power requirements are much more modest.

B. Spring Design Case #1

Again in figure 1, the peak slope of angular displacement coincides with a rapid decrease in ankle moment, plotting these two with respect to one another reveals a constant slope, or stiffness, relationship. Ideally a spring could be sized to this value and loaded slowly and earlier in the cycle, thus letting the power capabilities of the spring provide peak power for the gait cycle. This analogy is equivalent to setting motor power, P_m in equation 6, equal to zero at the instant of peak gait power, i.e 50% of the gait cycle. Solving this equation for K yields a stiffness value of 14152 N/m; thus for this example a 'power to weight' ratio for the spring becomes 3448 W/kg.[2]

C. Spring Design Case #2

Designing a spring to provide the peak power necessary for gait is an intuitive approach. Even though significant benefits can be gained by this approach the question of, "Is there a better stiffness?", can still be asked. To answer this question an exploration of the influence of stiffness, K, on peak motor power is required. Based upon equation 6, the relationship between stiffness, K, and 'peak' motor power is considered in equation 7.

$$(P_m)_{\text{peak}} = \max \left| F \cdot \dot{x}_g + \frac{F \cdot \dot{F}}{K} \right| \tag{7}$$

Again using the assumptions for our example person, an evaluation of equation 7 for a range of stiffnesses can be performed, see figure 3.

Figure 3 reveals an interesting relationship. Both extreme cases described earlier can be seen in this figure. At a stiffness of a value near zero, infinite motor power would be required. The analogy is that the spring is absorbing all of the power that the motor can provide and is not providing any back to the environment. At the opposite extreme is infinite stiffness or direct drive. It is seen that a high stiffness spring asymptotically approaches peak gait power near 250 W. However, rather than being a linear relationship between the two extremes, a minimum point or cusp occurs. The odd shape of this graph can be easily explained. The driving profile for this plot is determined by a $-\frac{1}{K}$ relationship with respect to power. The cusp is created as a function of the absolute value of this factor and hence a minimum is created. For the example problem, an optimal value of stiffness, K, is determined to be 20278 N/m.

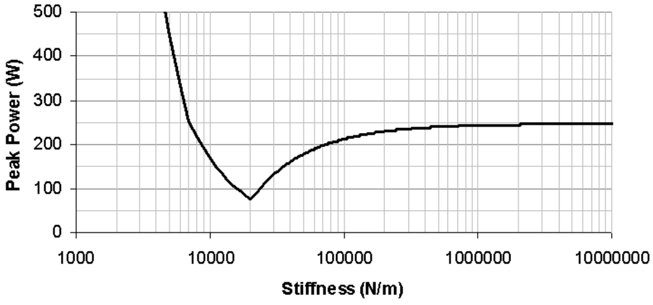

Fig. 3. Optimization of Stiffness, K.

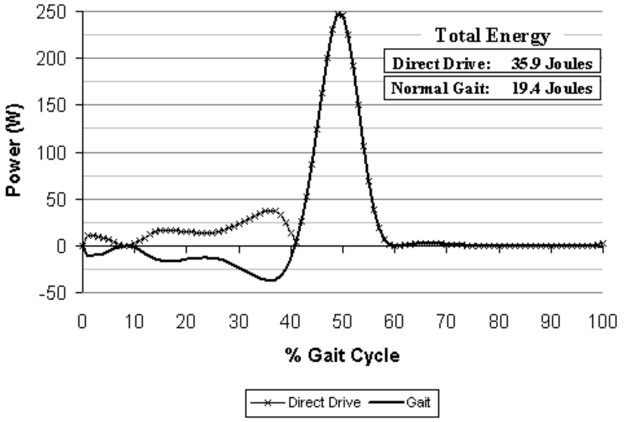

Fig. 4. Direct drive comparison to human ankle power in gait.

D. Ideal Comparisons

The motor power requirements that each of the scenarios describe are in fact 'ideal' situations, that is to say that no frictional effects are included. Considering friction and efficiency would of course increase actual power consumption. Nevertheless, considering each approach in terms of their own ideal requirements has great benefit. The first example compares a direct drive approach, or infinite stiffness spring, with the human ankle gait power requirements, see figure 4.

Notice that the peak power requirement that the direct drive system must provide is the same as the peak gait power requirement. Also notice that the negative power region for gait power is simply mirrored for a direct drive actuator. Again, a typical motor unit must provide braking for a negative power requirement. As a result of this mirroring, a much larger

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[2] Maximum power of a spring is $(P_s)_{\text{max}} = \frac{1}{2}K\Delta {x_s}^2\sqrt{\frac{K}{m}}$ (m is system mass) For our example, $(P_s)_{\text{max}} = \frac{1}{2}(14152\text{ N/m})(0.064\text{ m})^2\sqrt{\frac{14152\text{ N/m}}{1\text{ kg}}} \Rightarrow (P_s)_{\text{max}} = 3448\text{ W}$
If m is chosen to be the spring mass, then $(P_s)_{\text{max}} = 15420$ W, and the 'power to weight' ratio becomes 308000 W/kg.

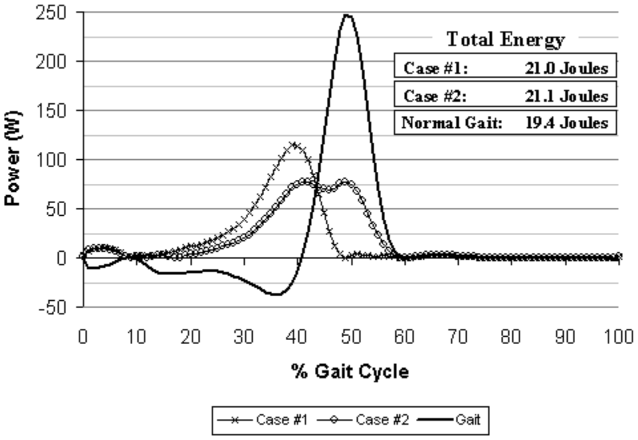

Fig. 5. Robotic Tendon comparison to human ankle power in gait.

amount of energy is required by the direct drive scenario (35.9 Joules) than is described for normal ankle gait (19.4 Joules). Where energy was calculated simply by integrating under the power requirement curve.

The second example compares the same human ankle data with that of the two cases for the Robotic Tendon design. Case #1 is the scenario in which spring stiffness is matched to 'push off' power requirements and case #2 is the scenario in which spring stiffness is selected based upon a minimum peak power. In either of these two cases the requirements for their respective peak power and energy are significantly reduced when compared to the direct drive example. Figure 5 illustrates the resulting powers obtained when applying the two stiffness cases.

Readily seen in both cases is that peak power is significantly reduced from what human ankle gait demands. This means that the spring is making up the differences in power. It is not hard to imagine that the human musculotendon structure may likely be doing something very similar. For case #1 peak power is reduced to 113 W and energy is 21.0 Joules. For case #2 peak power reaches a minimal value of only 77 W and the energy is just 21.1 Joules. Peak power of case #2 is only 31% of that required by the direct drive system, a comparable DC motor for this range of power would be the Maxon RE35, which is nominally rated for 90 W(rated peak power, 206 W) and weighs 0.34 kg (11% of the direct drive example), an applicable gearbox would likely scale as well. In addition, for each case the energy requirements are nearly half of those needed by the direct drive example. The power, weight and energy savings translate favorably to requirements demanded by wearable robotic systems.

V. ACTUATOR DESIGN

Armed with knowledge of appropriate spring selection methodology, a device to power human ankle gait can be constructed. However, unlike the idealized cases presented above, a real world design must consider device efficiencies and thus friction. Another consideration is that in humans, high power is accomplished by large torques and low velocities. However, a DC motor performs best at high velocities and low torques. In order to match human performances a gearbox/speed reducer must be added.

This can be a daunting prospect considering that the example gearbox discussed earlier weighed more than the motor itself and only had a maximum efficiency of $\eta = 0.7$. Fortunately, to a designer of wearable robotic devices, gearboxes are not the only means by which mechanical power can be converted. A DC motor can also be coupled to a lead screw or ball screw drive system, which can convert motor power to a form appropriate for human movement manipulation.

For reasons of cost and size preference, a lead screw design process similar to that described by Hollander and Sugar[18] was used. An often overlooked aspect of lead screw mechanism design is that both radius and pitch angle are required to effect lead. In a design requiring large mechanical advantage or a fine lead, low pitch angles ($< 10^o$) are often chosen. The downside to this approach is that excellent coefficients of friction are necessary to get good mechanical efficiency, i.e. a ball screw with $\mu \approx 0.003$ is need for mechanical efficiencies on the order of $\eta = 0.9$. In a contrasting approach, using a small screw radius and high pitch angles will accomplish an identical lead, but a only coefficient of friction of $\mu = 0.05$ is required to achieve mechanical efficiency of $\eta = 0.9$.

In order to meet the lead or mechanical advantage requirements and still use a high lead angle, an appropriately small radius must be chosen. Within the constraints of stress and force, a small radius screw has an advantage of very low weight. Weight in a screw decreases by the square of its radius. Lead requirement is determined by the system, but radius and pitch angle should be manipulated to minimize weight and maximize efficiency. For these reasons, the first prototype of the Robotic Tendon actuator features a small diameter lead screw ($\varnothing$6.4 mm) with a pitch angle of 23^o. To improve the performance of the commercially available lead screw, the addition of a nickel-teflon coating was used to provide a coefficient of friction with a value of $\mu = 0.05$, thus increasing the overall mechanical efficiency to a value of $\eta = 0.88$.

Although the lead screw efficiency is in the range we desired, the lead and radius are larger than would be preferred. This is simply due to construction timing and what lead screws are readily available. Rather than run the motor at low efficiency and low speed, a small 4.3:1 gearbox was added instead. The advantage is that the motor and gearbox can now be operated at a combined mechanical efficiency of $\eta = 0.81$, as opposed to the motor alone running at $\eta \approx 0.2$. The disadvantage is a 50% increase to the weight of the motor alone. The combined mechanism, a Maxon RE40 DC motor, a Maxon GP42C gearbox, a high efficiency lead screw and a peak power minimizing spring can be seen in the figure 6.

The Robotic Tendon actuator can provide the 250 W of peak power required by the ankle gait cycle in a package that weighs 0.95 kg, i.e. 'power to weight' ratio of 263 W/kg and is 0.366 m long extended (Note that the spring and the motor

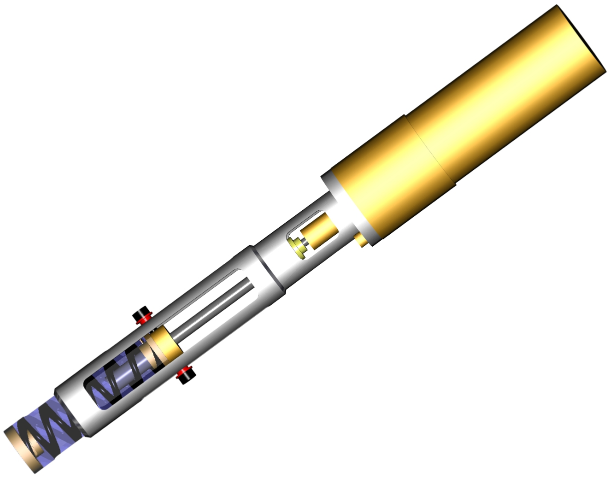

Fig. 6. Designed Robotic Tendon Actuator.

combine to achieve 250 W). In a future design, the gearbox can be removed thus improving the 'power to weight' ratio to 362 W/kg and its extended length to 0.325 m. Considering the efficiency of this actuator, the energy required by this device is 30 Joules per single leg step. As a comparison, the direct drive example detailed earlier can also provide the 250 W of peak power for the ankle gait cycle at a weight of 6.6 kg, i.e. 'power to weight' ratio of 38 W/kg. Also, considering the efficiency of this direct drive approach ($\eta = 0.63$) the energy require for each single leg step would be 57 Joules. The bottom line is that a Robotic Tendon actuator can cut energy requirements in *half* compared to a direct drive approach and can reduce peak power needs from its motor by a factor of *three*.

VI. CONCLUSIONS

To develop wearable robotic systems for human force and performance augmentation, devices that are portable, lightweight, safe, economical and simple to operate are required. Stated previously these demands can be difficult to achieve using traditional direct drive approaches. Again, in our simple example for ankle gait a DC motor/gearbox combination would require a minimum of 250 W of peak power and consume nearly 60 Joules of energy for each step. With the addition of 6.6 kg per ankle joint, this proposed direct drive scenario is neither lightweight nor portable for very long periods of time.

In contrast, a Robotic Tendon based device can do the same job with 7 times less weight and consume only half of the energy required by the direct drive example. In addition, use of a compliant spring adds a level of inherent safety in its attachment to a human operator. Compliance or give of a spring can help minimize the danger or damage a robotic instrument can do to a user in the case of a mishap. Also, including a lower power motor is inherently safer as well. Lastly, as the demands of the most complicated portions of ankle gait are handled via a spring, a simple and economical position controller scheme is possible. Currently, development of a Robotic Tendon with a variable stiffness spring is being pursued to further reduce the energy requirements.

A Robotic Tendon actuator allows a wearable assistive device to become a more practical reality to those in need of aid. With the significant benefits shown here, a Robotic Tendon actuator can help make wearable robotics a more prevalent part of our near future.

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