New horizons for orthotic and prosthetic technology: artificial muscle for ambulation

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ABSTRACT

The rehabilitation community is at the threshold of a new age in which orthotic and prosthetic devices will no longer be separate, lifeless mechanisms, but intimate extensions of the human body—structurally, neurologically, and dynamically. In this paper we discuss scientific and technological advances that promise to accelerate the merging of body and machine, including the development of actuator technologies that behave like muscle and control methodologies that exploit principles of biological movement. We present a state-of-the-art device for leg rehabilitation: a powered ankle-foot orthosis for stroke, cerebral palsy, or multiple sclerosis patients. The device employs a force-controllable actuator and a biomimetic control scheme that automatically modulates ankle impedance and motive torque to satisfy patient-specific gait requirements. Although the device has some clinical benefits, problems still remain. The force-controllable actuator comprises an electric motor and a mechanical transmission, resulting in a heavy, bulky, and noisy mechanism. As a resolution of this difficulty, we argue that electroactive polymer-based artificial muscle technologies may offer considerable advantages to the physically challenged, allowing for joint impedance and motive force controllability, noise-free operation, and anthropomorphic device morphologies.

Key words: human rehabilitation, orthosis, prosthesis, artificial muscle.

1. INTRODUCTION: THE STATE OF EXTERNAL ORTHOTIC AND PROSTHETIC LEG SYSTEMS

External, lower-extremity orthotic and prosthetic (O&P) devices have been around for centuries, if not millennia. However, until very recently, there were few attempts at applying engineering rigor and new technologies to improve mechanism function.[1] External O&P devices were made from crude materials such as wood and leather, making them heavy, nonadaptive, and difficult to use. In the 1970s all this began to change. Professor Woodie Flowers at MIT conducted research to advance the prosthetic knee joint from a passive, nonadaptive mechanism to an active device with variable damping capabilities.[2–5] Using the Flowers knee, the amputee experienced a wide range of knee damping throughout a single walking step. During ground contact, high knee damping inhibited knee buckling, and low damping throughout the swing phase allowed the prosthesis to swing freely before damping was increased to smoothly decelerate the prosthesis prior to heel strike. Unfortunately, the Flowers knee was never sold commercially. However, several O&P companies manufactured variable-damper knee products similar to the Flowers knee.[6] Actively controlled knee dampers offer advantages over mechanically passive knee systems. Most notably, transfemoral amputees can change gait speed and descend inclines and stairs with greater ease and stability.[7–10] Orthoses differ from prostheses in that they augment, rather than replace, the natural limb. Within the realm of orthotics, manufacturers recently developed a computer-controlled knee orthosis comprising a variable-clutch mechanism. Unlike the variable-damper prosthetic knees, the adaptive knee orthosis cannot vary knee damping but offers only locking and unlocking controllability. Even so, in comparison to conventional locked-knee designs, the adaptive knee orthosis improves the metabolic economy of level ground ambulation.[11]

In contrast to O&P knee devices, commercially available ankle-foot mechanisms are completely passive and nonadaptive. Today's prosthetic ankle-foot systems typically use elastomeric bumper springs or carbon composite leaf springs that store and release energy throughout each walking or running step. Compared to noncompliant or dissipative (damping only) ankle-foot devices, these contemporary elastic mechanisms offer considerable heel, toe, and vertical compliance to the below-knee amputee, increasing the amputee's perceived comfort and walking speed.[6] As with ankle prostheses, state-of-the-art ankle-foot orthoses are also passive, energy-storing devices. While traditionally ankle-foot

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orthoses have been constructed using rigid thermoplastics lined with soft foam, more contemporary designs are substituting these materials for lightweight, energy-storing composites.

Although tremendous technological progress has been made since the days of the wooden peg leg, contemporary O&P limbs cannot yet perform as well as their biological counterparts, whether in terms of stability, power generation, efficiency or cycle life.[12] However, in the next several decades, continued advances in muscle-like actuators and biomimetic control schemes may result in dramatic improvements in the quality of life of the physically challenged. In this paper, we describe the difficulties experienced by today's physically challenged due to limitations in current technology. We argue that O&P actuators must control joint impedance and motive force if artificial appendages are to truly mimic biological function, even during level ground ambulation. To demonstrate the clinical importance of powered O&P leg technology, we evaluate, by way of case study, the clinical efficacy of a powered ankle-foot orthosis. Finally, we make the case that artificial muscle technologies will offer considerable advantages to the physically challenged compared to traditional motor strategies, allowing for joint impedance and motive force controllability, noise-free operation, and anthropomorphic O&P device morphologies.

2. STATEMENT OF THE PROBLEM

The fact that lower-extremity O&P devices cannot modulate spring stiffness and motive force causes many problems for the physically challenged. One problem is balance. It is not uncommon for people suffering from leg disability to fall, especially while trying to walk over irregular surfaces. Recent studies suggest that balance would improve if O&P ankle stiffness were adjusted in response to changes in substrate mechanics. In 1997, researchers at Berkeley discovered that overall leg stiffness is increased to offset decreases in ground stiffness by normal individuals who hop and run.[13,14] Using a mathematical model to simulate leg function, they discovered that changes in knee and hip stiffness had little effect on overall leg stiffness, but changing ankle stiffness by 1.75-fold caused leg stiffness to increase 1.7-fold, suggesting that ankle impedance control was an important mechanism for the observed leg stiffness variations. Although it is not yet clear why the human ankle changes stiffness, researchers speculate that ankle impedance is modulated to optimize the smoothness and stability of bipedal ambulation.

In addition to problems of balance, people suffering from leg disability often tire more easily. For example, studies measuring the walking metabolism of transtibial (below-knee) and transfemoral (above-knee) amputees report a 40% and 60% increase in metabolic rate compared to normals, respectively. [15–17] Winter and Sienko [1988] hypothesized that the primary cause of the global transtibial metabolic increase is insufficient prosthetic ankle power generation.[18] At fast walking speeds, the human ankle generates a greater amount of positive work than negative work throughout each stance period.[1] In distinction, today's O&P ankles are passive springs, and therefore cannot generate more mechanical power than is absorbed throughout a gait cycle.[6] For transtibial amputees, Winter and Sienko [1988] observed a decrease in movement pathology when energy-storing prosthetic ankles were used instead of inflexible ankles, suggesting that additional increases in ankle power generation may normalize motor patterns and gait metabolism.[18] For transfemoral amputees, deficiencies in both ankle and knee power generation contribute to the relatively high gait metabolism. Although some improvements in gait have been observed with variable-damper knee designs, problems still remain. Variable-damper knees offer little to no improvement in gait metabolism compared to mechanically passive knees.[19] Popovic´ and Schwirtlich [1988] demonstrated that amputees using a powered knee prosthesis could walk at faster speeds and with an improved metabolic economy compared to conventional knees that only dissipate mechanical energy.[20] Although dramatic clinical advantages were achieved, their knee design was never commercialized because of problems with the mechanical system, including an inadequate cycle life, excessive mechanism noise, and a limited battery life.[21]

3. CASE STUDY: A POWERED ANKLE-FOOT ORTHOSIS

For artificial appendages to truly mimic biological function, even during level ground ambulation, O&P actuators must control both joint impedance and motive force. To demonstrate the clinical importance of powered O&P leg technology, we evaluate, by way of case study, the clinical efficacy of a powered ankle-foot orthosis. The orthosis, shown in Fig. 1, is designed to correct the dominant complications of drop-foot, a gait pathology most commonly caused by stroke, cerebral palsy, multiple sclerosis or traumatic injury. Drop-foot results from a particular muscle impairment in the anterior compartment of the leg where a patient is unable to dorsiflex the ankle or lift the foot. In walking, the major complications of drop-foot are 1) slapping of the forefoot after heel strike and 2) dragging of the toes at the beginning of each swing phase. The powered orthosis employs a force-controllable actuator and a control algorithm based on biomechanical models of normal ankle function.[22–24] Attached posteriorly to the ankle-foot orthosis is an actuator comprising a spring placed in series with an electric motor like a tendon in series with a muscle.[25] This series elasticity enables the system controller to modulate force instead of position. For this spring-plus-motor system, output force is proportional to the position difference across the series elasticity multiplied by the spring constant. By applying a position control on the spring, force or torque can be controlled across the orthotic joint.

The powered orthosis uses local mechanical sensors as an indirect measure of user intent. The external sensors, shown in Fig. 1, measure ankle joint position (potentiometer 3) and ground reaction force (capacitive force sensors 4). These sensory data are then used to control ankle impedance during stance and ankle position during the swing phase of walking. To alleviate the gait complication of drop-foot, the controller commands a linear spring law during controlled plantarflexion, from heel strike to midstance, where the actuator applies a torque proportional to joint position. As is shown in Fig. 2, this linear spring control makes the ankle function like a passive spring and is consistent with the response of a normal human ankle during level ground walking.[26] Although normal ankle mechanics can be characterized by a linear torsional spring, the stiffness of the ankle is actively modulated by the body from step to step, even at steady walking speeds. For the orthotic device, we find that when the applied ankle stiffness is too low, excessive forefoot collisions occur, causing the drop-foot condition. To alleviate this complication, the orthotic controller increases stiffness at each walking speed and for each terrain until forefoot collision forces are minimized. Throughout late stance, the controller minimizes joint impedance so as not to impede powered plantar flexion movements, and during the swing phase, the controller adjusts the position of the foot, dorsiflexing the ankle to provide sufficient toe clearance.

To assess the clinical performance of the powered orthosis, kinetic and kinematic gait data were collected on two drop-foot participants wearing both the powered orthosis and a conventional, constant spring stiffness orthosis.[23] These data were then compared to the gait mechanics of three age-, weight-, and height-matched normals. The preliminary data showed that the powered orthosis wearers more closely conformed to the normal subjects. During the stance period, the variable-impedance controller reduced the occurrence of slap foot and allowed greater late-stance, powered plantarflexion movements. In addition, during the swing phase, the position control applied by the orthosis to dorsiflex

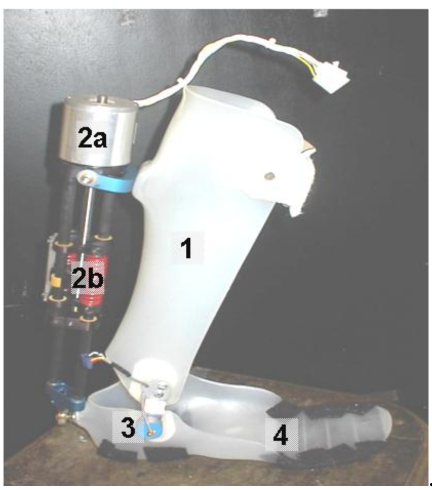

Fig. 1. A powered ankle-foot orthosis. Using a series-elastic actuator, the stiffness of the ankle joint is modulated from step to step to control the movement of the foot during controlled plantarflexion. In addition, the motor system dorsiflexes the foot during the swing phase to achieve foot clearance. The ankle-foot orthosis (1) comprises a series-elastic actuator that is composed of a motor and ball screw (2a) and series elasticity (2b), potentiometer angle sensor (3), and capacitive force sensors (4).

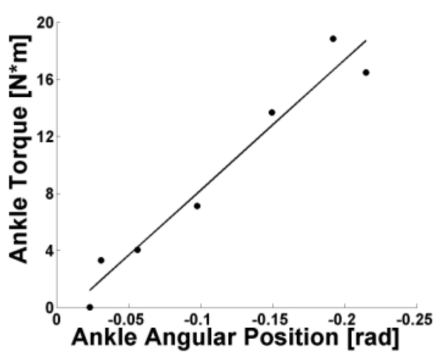

Fig. 2. Ankle torque versus position data for a normal subject walking on a horizontal surface. Only ankle data during the controlled plantarflexion phase of walking are shown. Although data for just one subject and one walking step are plotted, the human ankle behaves like a linear spring throughout early stance independent of walking speed, storing and releasing energy, and allowing for a smooth heel to forefoot strike sequence.[26]

the ankle provided for less kinematic difference when compared to normals. These results indicate that a powered orthosis that actively modulates both impedance and motive torque may have certain clinical benefits for the treatment of drop-foot gait compared to conventional ankle-foot orthoses having constant stiffness joint behaviors.

Although some improvements in gait were achieved with the powered orthosis, design difficulties still remain. The powered orthosis is unnecessarily heavy, making walking more difficult. Still further, the device is both unnatural in shape and noisy, factors that negatively effect mechanism cosmesis and comfort. In the next section, we argue that electroactive polymer artificial muscle technologies can overcome these limitations and may offer considerable advantages to the physically challenged, allowing for integrated joint impedance and motive force controllability, noise-free operation, and anthropomorphic device morphologies.

4. RESOLUTION OF THE DIFFICULTY: MUSCLE-LIKE ACTUATION

The actuator and transmission play a dominant role in determining the dynamic performance of O&P devices. Muscles make up a significant portion of a biological limb's mass and also determine its general shape. In trying to duplicate the form and function of a limb it therefore makes sense to use actuators that are similar to natural muscle. While it is not possible to duplicate natural muscle in every respect, we do need to match those characteristics of natural muscle that provide for robust and adaptive locomotion as well as those factors that affect cosmesis and comfort. The preceding sections describe the importance of duplicating certain key features of natural motion about a joint. In particular, natural muscle is more than just a motor (force and motion producer). Muscle is also an energy storage device (spring) and an energy absorber (damper). In addition to optimizing the metabolic economy of walking, the energy storage and absorption characteristics (and the ability to modulate those parameters) help a person respond to unknown disturbances, much like the suspension system of an automobile. All these functions are integrated to produce the desirable characteristics of natural walking or other gaits. Strictly speaking, some of the elasticity and damping of natural muscle is borne by the tendons that attach the muscles to the skeleton. For convenience, we will talk about "muscle" as including the function of the tendons. Beyond the stiffness, damping, and force and stroke output capabilities, as well as the cosmetic and comfort issues, it is important to note that biologists and engineers do not yet know all the features of muscle that might be required for natural motion. For example, muscle has very nonlinear time-varying behavior that could be important. However, at a minimum we should be able to reproduce the power output, force, stroke, elasticity, and damping (or energy absorption) of natural muscle.

While in theory a suitably fast and powerful actuator acting through a mechanical transmission could use closed-loop feedback control to imitate the performance of natural muscle, such as the series elastic actuator described in the previous section, actuators that share the inherent properties of muscle can be lighter, simpler and offer a more natural look and feel. The requirement for a "natural" look and feel is important to the goal of making the artificial or assisted limb feel integrated with the body. Such actuator requirements are not typically considered in robotic or dynamic machinery applications when actuators are selected. These requirements include completely quiet operation, soft feel, and an external envelope (size and shape) approximating that of a natural limb. In addition, cost is very important. The cost of some shape memory alloys, piezoelectric, and magnetostrictive materials could prevent their use in applications such as lower limbs where large amounts of material are required.

Many actuator materials and devices have been put forth as "artificial muscles." Pneumatics and hydraulics (including soft-bodied actuators such as the "McKibben Muscle" can imitate much of the performance of natural muscle and have a shape and feel similar to natural muscle, but they are noisy, difficult to control, and require a separate pump to provide the fluid energy.[27] Most typically it is desired to use actuators that can be powered and controlled directly by electricity. Table 1 compares the performance of electrically controllable actuation technologies with that of natural muscle. Existing technologies are all lacking in some way. Shape memory alloys, while strong with high energy density, are slow and inefficient. Piezoelectrics are fast and efficient but are stiff with a low peak strain. Electromagnetic motors, by far the most common actuator in existing robotic, prosthetic, and orthotic devices, do not offer an adequately high peak or average power output, resulting in devices that are heavier than desired. Furthermore, electric motors achieve maximum power output at high speed and therefore require a high-ratio, nonbackdrivable gearbox or ballscrew in order to produce the speeds needed for limb motion. Since a highly geared system cannot reproduce the compliance of natural muscle, it is necessary to include yet additional elements to provide

Table 1. Comparison of candidate actuator technologies and natural muscle
(adapted from Kornbluh et al. 2004 and Madden et al. 2004) [28,29]

	Actuator Type (specific example)	Typical (Max.) Strain (%)	Typical (Max.) Stress (MPa)	Typical (Max.) Specific Elastic Energy Density (J/g)	Typical (Max.) Elastic Energy Density (J/cm^3)	Typical (Max.) Avg. Specific Power Density at 1 Hz (W/g)	Peak Strain rate (%/s)	Elastic Modulus (MPa)	Est. Max. Efficiency (%)	Relative Speed (full cycle)
NATURAL MUSCLE	Mammalian Skeletal Muscle	20 (40)	0.1 (0.35)	0.041 (0.08)	0.041 (0.08)	0.041 (0.08)	> 50	10–60	20%	Medium
ELECTROACTIVE POLYMER	Dielectric elastomer	25 (> 300)	1.0 (7.0)	0.1 (3.4)	0.1 (3.4)	0.1 (3.4)	> 450	0.1–10	60–90	Med. - Fast
	Electrostrictive Polymer	3.5 (7.0)	20 (45)	0.17 (> 0.53)	0.3 (> 1.0)	0.17 (> 0.53)	> 2000	400–1200	60–90	Fast
	Electrochemo-mechanical Conducting Polymer	2 (20)	5 (200)	0.1 (1.0)	0.1 (1.0)	0.1 (1.0)	1	200–3000	< 5	Med. -Slow
	Ionic Polymer Metal Composite	0.5 (3.3)	3 (15)	(0.004)	(0.006)	0.004	3.3	50–100	1.5–3	Med. - Slow
	Mechano-chemical Polymer/Gels (Polyelectrolyte)	> 40	0.3	0.06	0.06	< 0.06	< 1	?	30	Slow
	Piezoelectric Polymer (PVDF)	0.1	4.8	0.0013	0.0024	0.0013	?	450	60–90	Fast
OTHER	Liquid Crystal Elastomer (Thermal)	19 (45)	0.12 (0.45)	0.003 (0.06)	0.003 (0.06)	< 0.003	37	0.3 – 4	< 5%	Slow
	Shape Memory Polymer	100	4	2	2	< 0.2	?	?	< 10	Slow
NONPOLYMER ACTUATORS	Electromagnetic									
	Direct (Voice Coil)	50	0.10	0.003	0.025	0.003	> 1000	NA	> 80	Fast
	Motor/transmission	50	NA	NA	NA	0.5	< 200	NA	> 50	Medium
	Piezoelectric									
	Ceramic (PZT)	(0.2)	(110)	(0.013)	(0.10)	(0.013)	> 1000	25,000–70,000	> 90	Fast
	Single Crystal (PZN-PT)	(1.7)	(131)	(0.13)	(1.0)	(0.13)	> 1000	9000	> 90	Fast
	Shape Memory Alloy (TiNi)	> 5	> 200	> 15	> 100	< 15	300 (one direction only)	20,000–80,000	< 10	Slow
	Thermal (Expansion)	1	78	0.15	0.4	< 0.15	Depends on heat transfer	> 70,000 (varies)	< 10	Slow
	Magnetostrictive (Terfenol-D, Etrema Products)	0.2	70	0.0027	0.025	>0.0027	>1000	40,000	60	Fast

for muscle-like features, such as the series spring in the series-elastic actuator described earlier. While the functionality of the resulting actuator is good, the mass and unnatural noise of the transmission mechanism are unattractive. For orthoses in particular where the mass of the natural limb remains, minimizing the excess mass as well as bulk of the actuator is critical.

Since natural muscle is basically a polymer, it is not surprising that polymer-based artificial muscle behaves most like natural muscle. Indeed, the term *artificial muscle* is often used interchangeably with polymer actuators. There are many types of polymer actuators. Most fall into one of four categories 1) mechanochemical gels, 2) electrochemical (wet), 3) field-activated (dry), and 4) thermally activated or phase change. A summary of types of polymer actuators can be found in Bar-Cohen [2001], Sommer-Larsen and Kornbluh [2002], or Madden et al. [2004].[29–31] Each type has shown some promise in artificial muscle applications, although many have significant limitations. Compared to most other materials, polymer actuators are very low cost.

A wide variety of gel actuators have been developed.[32] Fundamentally, all are chemically responsive but some can use electricity to control the amount of ionic species that, in turn, control swelling. In a clear demonstration of the use of electroactive gels in lower limbs, Shahinpoor and others demonstrated electrically controllable mechanochemical gel fibers contracting artificial muscles attached to the legs of a life-size model of a human skeleton.[33] This powered skeleton, named “Myster Bony,” was able to slowly pedal a bicycle. However, the response speed and power density of gels is presently too low for normal walking. Gels are fundamentally diffusion limited (they typically rely on the transport of fluids and ions), so increases in speed and power require changes in the microstructure of the gel-plus-fluid delivery system. In biomimetic terms, they need a better circulatory system. Some gels can be powered by organic solvents much in the same way that muscles are powered by nutrients. Thus, artificial muscles might be powered directly by fuels. The chemical-to-mechanical energy conversion efficiency of natural muscle is about 20%—comparable to many combustion engines.

Electrochemical polymers include both conducting polymers and ionic polymer metal composites.[34,35] They too are largely diffusion limited and so are relatively slow and difficult to scale to large sizes. Further, the electromechanical coupling of electrochemical muscles is very low, suggesting that it is difficult to achieve the high efficiency needed for O&P applications. The strain of electrochemical muscles is typically a few percent but can be as high as 10%.

Thermally responsive polymers include shape memory polymers and phase change materials such as liquid crystal elastomers.[36,37] Such materials are capable of undergoing large strain changes, but their response speed is typically slow as it is heat transfer limited and overall energy efficiency is very low.

Field-activated polymer muscles include electrostrictive polymers that undergo a shape change due to realignment of a crystalline phase in the presence of an electric field as well as dielectric elastomers comprising rubbery insulators that undergo a shape change due to the electrostatic forces of the charges on their compliant electrodes.[38,39] The strain of electrostrictive polymers can be as high as 5%. The strain of dielectric elastomers can exceed 100%—duplicating that of natural muscle. Field-activated polymers have several other properties that make them attractive for O&P leg systems. Field-activated polymers can respond quickly and have high electromechanical coupling that can allow for overall efficiencies as high as 80%. Perhaps most significantly, they can exceed the peak power of natural muscle, allowing for devices of size and mass comparable to natural leg muscles.

The stiffness of dielectric elastomers is similar to that of natural muscle, but the stiffness of electrostrictive materials is more than an order of magnitude greater.[40] Since field-activated polymers have a fast response, the force and strain of these materials could be electrically modulated to control stiffness.[28] Actuators that already have a stiffness in a desired operating range are easier to control and more robust since such actuators can more easily respond to high-frequency shocks or disturbances that may be difficult to handle using feedback. Inherently compliant actuators can also be used in antagonistic pairs to modulate joint stiffness over a wider range than is possible with a single actuator.

Since field-activated polymers have high electromechanical coupling, they can also be operated in reverse as a generator. Thus, they can be used in a semi-active mode to provide damping or energy absorption (without the need for additional damping components). This mode of operation could also be used to recapture some of the electrical energy applied to the muscle. It is also possible to use soft viscoelastic polymers in parallel with the actuators to add more passive damping.

Dielectric elastomer actuators have already shown some promise as artificial muscles for O&P devices. These actuators have been formed into cylindrical rolls that have strain, shape, and performance similar to natural skeletal muscles.[41] Such actuators have been demonstrated acting as biceps on a life-size skeletal arm, although the particular actuator was smaller than would finally be desired.[42] Figure 3 shows a photo of this device. These smaller actuators can be grouped to increase the stroke and force, much as individual muscles are composed of parallel fibers that in turn are made up of series contractile units.

Fig. 3. Rolled dielectric elastomer acting as a bicep on a full-size human skeletal muscle. Field-activated polymers operate at high voltage and low current. Further development of lightweight and efficient voltage conversion and driving circuitry is required for operation off of batteries. High voltage must be isolated from the user. However, the potential danger of high voltage can be greatly reduced by limiting the available current. For example, operation of 100 W actuators at 5000 V (a typical maximum for dielectric elastomers) would require just 20 mA —a nonlethal amount.

5. CONCLUDING REMARKS

Although tremendous technological progress has been made in the area of permanent assistive devices, people suffering from leg disability still experience balance difficulties and high walking metabolisms. As a resolution to these difficulties, we feel that continued advancement in muscle-like actuators is of critical importance. In particular, electroactive polymers have shown considerable promise as artificial muscles but technical challenges still remain in their implementation. These challenges include improving the actuator's durability and lifetime at high levels of performance, scaling up the actuator size to meet the force and stroke needs of O&P devices, and advancing efficient and compact driving electronics. Although difficulties remain, electroactive polymers have the potential to restore natural motion to the physically challenged through the active modulation of artificial joint impedance and motive force in the context of a quiet, low-mass, and morphologically realistic artificial appendage.

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