CHAPTER 32
DESIGN OF ARTIFICIAL ARMS AND HANDS FOR PROSTHETIC APPLICATIONS

Richard F. ff. Weir, Ph.D.,
Northwestern University Prosthetics Research Laboratory, and
Rehabilitation Engineering Research Center, Chicago, Illinois

32.1 INTRODUCTION
The design of fully functioning artificial arms and hand replacements with physiological speeds-of-response and strength (or better) that can be controlled almost without thought is the goal of upper-extremity prosthetics research. Unfortunately, current prosthetic components and interface techniques are still a long way from realizing this goal. The current state-of-the-art prosthesis can be considered to be a tool rather than a limb replacement.

The prosthesis as a tool makes no pretense of trying to replace the lost limb physiologically but is there as an aid to help provide some of the functions that were lost. The prosthesis as a tool is an interchangeable device that is worn and used as needed, and then ignored. Much effort in the field of upper-extremity prosthesis research is directed toward the creation of prostheses as true limb replacements; however, in current practice we are mostly limited to prostheses as tools.

The major factors limiting prostheses to tools are practical ones due to the severe weight, power, and size constraints of hand/arm systems as well as the difficulty in finding a sufficient number of appropriate control sources to control the requisite number of degrees of freedom. Of these, it is the lack of independent control sources that imposes the most severe impediment to the development of today’s prosthetic hand/arm systems. As a result, upper-limb prosthetics research is somewhat dominated by considerations of control. Still, the importance of better actuators and better multifunctional mechanisms must not be ignored. Control is useless if effective hand and arm mechanisms are not available.

The problems associated with the design of artificial hand and arm replacements are far more challenging than those associated with the design of robotic arms or terminal devices. In fact, robotics and prosthetics design has much less in common than one might expect. Robotics concepts have had little impact on commercial prosthetics because of the significantly different physical constraints required for a prosthetic device to be successful. Although some size, weight, and power constraints
must be placed on robots and manipulators, robotic actuators can often be as large and as heavy as
required to achieve a specific result. Power is usually not an issue since it can be obtained from the
power mains. Prosthetic arm and hand design can be viewed as a subset of the greater field of robot
and manipulator arm and end-effector design.

Robot arms look impressive. However, announcements by robot arm designers who, when
searching for an application for their new mechanical arms, claim their new robot will be a boon to
the field of prosthetics but should be treated with skepticism. The issue has never been about an
inability to build mechanical arms and hands. The MIT/Utah dexterous hand (Jacobsen et al., 1984)
is an example of a mechanical hand that mimics the function of a hand. This hand was designed for
use in research studying robot dexterity. This device could never be used in prosthetics because the
actuators and computer system required to control this hand occupy the space of two small filing
cabinets, and power is supplied externally from electrical mains. The real issue in upper-limb
prosthetics, of which most robot arm designers seem to be unaware, is “How does one interface this
arm to the person?” and “How is the arm to be controlled?”

Current state-of-the-art prosthetic hands are single-degree-of-freedom (opening and closing)
devices often implemented with myoelectric control (a form of control that uses the electric fields
generated as a by-product of normal muscle contraction). Prosthetic arms requiring multi-degree-of-
freedom control most often use sequential control with locking mechanisms to switch control from
one degree of freedom to another. Generally, since vision is the primary source of feedback, the
greatest number of functions that are controlled in parallel is two. Otherwise, the mental loading
becomes excessive. Switch, myoelectric, or harness and cable are the primary modes of control for
today’s upper-limb prostheses.

Eugene F. Murphy, Ph.D., (1955) a former chief of the Research & Development Division,
Prosthetic & Sensory Aids Service, Veterans Administration (Hays, 2001), probably articulated best
the awe and frustration associated with the task of trying to replicate the function of the natural hand
by mechanical means, when he wrote in “Engineering—Hope of the Handless”:

The human hand, with its elaborate control system in the brain, is doubtless the most widely versatile machine
that has ever existed anywhere. Its notorious deficiency lies in its persistent inability to create a similar
machine as versatile as itself. This circumstance accounts for the fact that, while there has been from earliest
times a great need for hand replacements, all attempts to produce successful hand substitutes have thus far
ended in only a rather crude imitation of a very few of the many attributes of the living counterpart. For want
of complete knowledge of the natural hand-brain complex, and of the ingenuity requisite even to the most
modest simulation of the normal hand, artificial hands have always resembled the natural model in a superficial
way only. Voltaire is said to have remarked that Newton, with all his science, did not know how his own hand
functioned.

The design of artificial arms and hands is a multidisciplinary endeavor. The design team needs an
understanding of the mechanics of mechanisms, such as gears, levers, and points of mechanical
advantage, and electromechanical design, such as switches, dc motors, and electronics. Mechatronics
is the new word used to describe this marriage of mechanical and electronic engineering. In addition
to mechatronics, the prosthetics designer must also have knowledge of musculoskeletal anatomy and
muscular- as well as neurophysiology.

It is the goal of this chapter to serve as a resource for designers of artificial limbs who come to the
problem with different areas of expertise. As such, I am compelled to refer the reader to the Atlas of
Limb Prosthetics (Bowker and Michael, 1992). This text is a comprehensive guide to the surgical,
prosthetic, and rehabilitation techniques employed in current clinical practice. From a historical
perspective, Human Limbs and Their Substitutes (Klopsteg and Wilson, 1954) is invaluable and
contains much biomechanical data on persons with amputation that is still valid. Consequently, both
these texts are referenced throughout this chapter. Also, following the philosophy of “Why reinvent
the wheel?” wherever possible, examples are provided of commercially available devices that use the
mechanisms, algorithms, or control approaches mentioned in the text.

It is hoped that the reader will find the information in this chapter to be practical and useful and
an aid in eliminating much of the startup time usually associated with familiarizing oneself with a new
topic. Ultimately, it is hoped that the information imparted will aid and facilitate in the design of better artificial arm and hand replacements for people with upper-limb amputations and deficiencies.

32.2 THE NATURE OF THE PROBLEM

There are over 30 muscles acting on the forearm and hand. The human hand has 27 major bones, and at least 18 joint articulations with 27 or more degrees of freedom (DOF). The arm contributes another 7 degrees of freedom. The primary role of the arm is to position the hand in space. The primary role of the hand is to enable a person to interact with the environment. Control of a person’s arm is directed at controlling the position of the arm’s hand. Even though people control their arms with great facility, this is a highly complex and demanding task.

A backhoe is essentially a mechanical arm that is under the control of an operator. To control this mechanical arm the operator uses both arms, both feet, both eyes, and all his or her concentration (Fig. 32.1). The driver uses both arms to pull levers, both feet to press pedals to operate the arm, and both eyes to monitor the task being performed by the digger—all this to control a single mechanical arm. Now consider a person with amputations of both arms above the elbow and one begins to have some appreciation of the task such a limbless person faces in controlling an artificial (prosthetic) arm or arms.

As for performance, the anatomical hand is capable of speeds in excess of 40 rad/s (2290 degrees/s), and grasps involving all fingers of the hand can exert up to about 400 N (90 lbf) of force. Average physiological speeds for every day pick-and-place tasks have been found to be in the range of 3 to 4 rad/s (172 to 200 degrees/s), while most activities of daily living (ADL) require prehension forces in the range 0 to 67 N (0 to 15 lbf) [these forces are dependent on the coefficient of friction between the gripping surface and the object held (Heckathorne, 1992)].

For the wrist and forearm normal ranges of motion (ROM) (Fig. 32.2, top) are 85 to 90 degrees of pronation, 85 to 90 degrees of supination, 15 degrees of radial deviation, 30 to 45 degrees of ulnar deviation; 80 to 90 degrees of wrist flexion, and 70 to 90 degrees of wrist extension (Magee, 1987). It has been found that for ADLs, 100 degrees of forearm rotation, 80 degrees of wrist flexion-extension, and 60 degrees of radial- ulnar deviation are sufficient (Heckathorne, 1992). The forearm can achieve maximum rotational velocities in excess of 14 rad/s (800 degrees/s) for pronations (inward rotations) and 20 rad/s (1150 degrees/s) for supinations (outward rotations).

For the elbow, normal range of motion (Fig. 32.2, bottom, left) is 140 to 150 degrees of flexion and 10–15 degrees of hyperextension. Peak anatomic elbow speeds of 261 degrees/s (4.5 rad/s) have been found (Doubler, 1982) for 90-degree movements, and the average male can generate an elbow torque of 138 N·m (102 ft·lbf) in flexion with the elbow at 90 degrees. In extension the average male can generate 75 percent of the maximum flexion torque.

For the upper arm, normal ranges of motion (Fig. 32.2, center) are 90 degrees of medial (inward, toward the midline) humeral rotation and 40 degrees of lateral (outward away from the midline) humeral rotation, 180 degrees of flexion (forward rotation of the arm about the shoulder) and 45 degrees of extension (backward rotation of the upper arm about the shoulder), and 180 degrees of elevation (abduction, outward rotation about the shoulder) and 20 degrees of depression (adduction, inward rotation of the upper arm about the shoulder).

For the shoulder complex, the normal ranges of motion (Fig. 32.2, bottom, center and right) are 40 degrees of elevation and 10 degrees of depression, 15 degrees of extension (scapular adduction), and 20 degrees of flexion (scapular abduction). Motion at the shoulder is not due to a single joint but is instead a combination of the motion associated with the scapulothoracic interface, the sternoclavicular joint, and the glenohumeral joint. The glenoid cavity in which the head of the humerus sits is made up of the acromion of the scapula, the spine of the scapula, and the clavicle. As such there is no single shoulder joint but rather a shoulder complex or girdle. The primary power source for body-powered prostheses uses a combination of glenohumeral flexion [forward rotation of the upper arm about the shoulder (glenohumeral joint)] and bicipital abduction (rounding of the
shoulders). This motion can result in excursions of up to 10 cm (4 in) with a force of 178 to 266 N (40 to 60 lbf). Knowledge of the normal ranges of motions for the arm and hand is important, particularly when designing body-powered systems. For more in-depth information, Sarrafian (1992) provides extensive anatomical range-of-motion data relating individual muscle contributions to different arm motions.

Another property of the physiologic arm is that it has “give,” that is, it is compliant or springlike. This compliance is not a fixed quantity but can be varied depending on the task requirements: a stiff arm for bracing oneself against an expected blow or a relaxed arm for playing the piano. This inherent compliance of the human arm also provides protection for the joints and musculo-skeletal system. Because the musculoskeletal system is compliant, it can withstand external shock loads far better than can a stiff-jointed equivalent.

Interaction with the real world is something current robotic and prosthetic actuators (dc electric motors with gear trains) do not do well. When a stiff robot arm comes into contact with a hard
FIGURE 32.2 Ranges of motion for the wrist (top), upper-arm (center), elbow (bottom-left), and shoulder (bottom, center and right). Normal ranges of motion are for the wrist and forearm, -85° to 90° of pronation, 85° to 90° of supination, 15° of radial deviation, 30° to 45° of ulnar deviation, 80° to 90° of wrist flexion, and 70° to 90° of wrist extension; for the elbow, 140° to 150° of flexion and 10° to 15° of hyperextension; for the upper arm, 90° of medial humeral rotation and 40° of lateral humeral rotation, 180° of flexion and 45° of extension, and 180° of abduction and 20° of adduction; for the shoulder, 40° of elevation and 10° of depression, 15° of scapular adduction and 20° of flexion. (Many thanks to Ms. Pinata Hungspreugs for her help in making this image).
surface, a phenomenon known as contact instability can arise. Unless robot, environment, and the
nature of the mechanical interaction between the two are precisely modeled mathematically, contact
instabilities can occur, even if the mathematical representation is only slightly in error. The human
arm does not have contact stability problems due to its inherent compliance.

The performance of current artificial mechanisms comes nowhere close to meeting the maximum
speed and force of which the anatomic arm and hand are capable, although hand mechanisms are
available that can attain speeds in excess of 3 rad/s and pinch forces in excess of 110 N (25 lbf).
These mechanisms, such as the NU-VA Synergetic Prehensor, (Hosmer-Dorrance, Calif.), the Otto
Bock Sensor and DMC hands (Otto Bock Healthcare, Duderstadt, Germany), and the Motion Control
Hand (Motion Control, Salt Lake City, Utah) can achieve this performance when overvoltaged as is
commonly done in clinical practice. As for wrist components, with the exception of the Otto Bock
Electric Wrist Rotator (Otto Bock Healthcare, Duderstadt, Germany) and the VASI (Varity Ability
Systems, Inc, Toronto, Canada) child-size electric wrist, all prosthetic wrist components are body-
powered and, when used, are used for positioning purposes. As such, these devices operate at low
speeds (less than 1 rad/s) with minimal torque-generating capability.

Current electric powered prosthetic elbows can attain about 12.2 N·m (9 ft·lbf) of “live-lift” (lift
by the elbows own motor mechanism) at speeds of about 2 rad/s (Boston Elbow III, Liberating
Technology, Inc., Mass.). Body-powered elbows are limited by the speed and strength of the user
and the efficiency of the linkage used to connect the user and the component. Humeral rotation for elbow
components, with the exception of the RIMJET body-powered humeral rotator (RIMJET, Fla.), is
achieved with manually positioned friction joints or turntables. The only shoulder joints available are
also passive, manually-positioned units that use friction or a lock to hold their position.

Thus it is apparent that while the user-prosthesis interface is a major impediment to the
advancement of prosthetic technology, there is much room for improvement in the prosthetic
components themselves. The limitations of current systems are not due to a lack of innovative design
but rather are due to the very severe nature of the physical constraints that are placed on the designer
and the inability of current technology to match the power and energy density of natural muscle.

### 32.3 GENERAL DESIGN CONSIDERATIONS

#### 32.3.1 Form versus Function

The role of form, or cosmesis, in prosthetics cannot be overstated. Often the design team will sacrifice
cosmetic appeal to achieve increased prehensile function. However, the relative importance of ap-
pearance versus function is highly dependent on the person with the amputation. Some people may
be solely concerned with visual presentation and reject a highly functional body-powered, cable-
operated prosthesis because of the unacceptable appearance of the control harness or of a hook-
shaped terminal device. Others might find the function provided by these devices sufficient to
outweigh their concerns about their appearance. Still others prefer a more toollike appearance believ-
ing a prosthesis that looks like a natural arm or hand but is not is a deception.

It is not uncommon for a person with an amputation to have two prostheses, one that emphasizes
mechanical function, one for work, and an interchangeable one with a more humanlike appearance for
social occasions, i.e., different tools for different jobs. Choice of prosthesis is ultimately based on many
psychological, cultural, and practical factors. Other factors affecting the issue are age, gender, occupation,
degree of physical activity, the amputee’s attitude toward training, the type of amputation involved, and
whether it is unilateral or bilateral limb loss. (Beasley and de Bese, 1990; Pillet and Mackin, 1992).

Cosmesis is the term used in prosthetics to describe how a particular device looks. A device is
considered to be cosmetic in appearance if it is aesthetically pleasing and looks like the limb it is
designed to replace in both its lines and color. However, a device may be statically “correct” in
appearance but it can look “wrong” or lifeless when it is in motion. In this instance, the device has
good static cosmesis but poor dynamic cosmesis.
People see what they expect to see, so if a person with an artificial limb replacement looks and moves in an expected manner the fact of the artificial limb will often go unnoticed by casual observers. Often a hand replacement can have only a vague resemblance to the natural hand but because it is moved in a natural-looking manner, it can pass undetected unless closely scrutinized. Here the device has poor static cosmesis but good dynamic cosmesis.

Dynamic cosmesis is frequently the more important of the two forms of cosmesis, but it is frequently overlooked because it is difficult to achieve. Dynamic cosmesis can be enhanced by preserving as much of the person’s residual motion as possible. For example, a partial hand prosthesis should not interfere with residual wrist motion because the wrist is used extensively in the positioning of the hand in space. Finally, a device can be considered to be functionally cosmetic if at a glance it is not immediately recognizable as an artificial hand regardless of whether it is in motion or not or whether it is or is not handlike when stationary.

Cosmetic gloves, made of poly vinyl chloride (PVC) or silicone, are generally employed to cover artificial hand mechanisms to give them good static cosmesis (Fig. 32.3). These coverings serve to increase the passive adaptability of a prosthetic hand to the shape of a grasped object and to increase the coefficient of friction of the hand. The price paid for this increased cosmesis and grasping function is a restricted range of motion and hindered performance (speed/force output) of the hand. This is so because the hand motors must overcome the elastic forces inherent in the glove. The ideal glove should enhance the cosmesis of the prosthesis without interfering with its performance.

Limb replacements should be anthropomorphic in general shape, size, and outline. This does not mean that an artificial hand or arm needs to look exactly like its human counterpart. However, there should be a joint that operates like an elbow joint where one would expect to see an elbow joint, and the various limb segments should be of a size consistent with a normal human being, i.e., any replacement should have similar kinematics and kinetics. With regard to the issue of hand size, artificial hands are usually selected to be smaller than their physiological counterparts. This is so because artificial hands are perceived to be larger than they really are, probably due to their rigid structure and essentially static appearance.

32.3.2 Weight

Final weight of a prosthesis is critical to the success of any prosthetic fitting. Contrary to what one might think, one should not make an artificial limb replacement the same weight as the limb it replaces. The weight of an adult male arm is about 10 kg (20 lb). Total arm replacements that exceed 3.5 kg (~7.5 lb) cannot be expected to be worn and used for a full day because of the discomfort associated with suspending that much weight from the body. Artificial arms need to be as light as possible or else they will end up in a closet. The lack of an intimate connection between amputee and limb replacement means that the prosthesis is perceived as an external load and therefore as something that must be carried. To be effective, artificial arms should be worn by their users for periods in excess of 8 to 12 hours a day. To gain some insight into how this might feel, consider carrying a 6-lb laptop computer slung from your shoulder for a day.
32.3.3 Power Sources

As is the case for all portable devices, power is scarce. Choice of power source defines a prosthesis, in that it determines the choice of actuator. If the power source is to be the person, i.e., body power, then the actuator is the person’s own musculature and the prosthesis should not require excessive effort to use. Mechanical mechanisms need to be efficient and frictional losses need to be minimized to avoid tiring the user over the course of a day. If the artificial limb is externally powered (i.e., uses a power source other than the body, usually electric storage), the limb should be able to run for a day from the same power source without needing to be replaced or recharged. In addition, it is desirable for the power source to be contained within the prosthesis.

Electrochemical batteries are the main source of energy for modern externally powered prosthetic arms, although pneumatic gas cylinders have been used in the past. There are a number of other technologies that could replace batteries as portable sources of electricity in the future. These include electromechanical flywheel systems that store energy in a rotating disk, and miniature Wankel type rotary combustion engines. However the most promising technology is that of ethanol- or methanol-based fuel cells. These devices are already moving into production for interim cell phone products. All are heavy and occupy excessive space. If electricity is the power source, then for the foreseeable future, dc electric motors will be the actuators. The problem of portable prosthesis power is analogous to the power issues in the laptop computer and cellular phone industry, where a major contributor to the weight and space of these portable devices is the power source, which for the moment is the battery. In addition, prostheses need high power density as well as high energy density.

32.3.4 Body-Powered Power Sources

In a body-powered device, the person uses his or her own muscular power to operate the prosthesis, usually via a cable link called a Bowden cable (Fig. 32.4). A Bowden cable consists of two parts, an outer housing and an inner tension cable. The housing is fixed at both ends and serves as a flexible bridge between two points, maintaining a constant length regardless of any motion. The cable is free to slide within the housing.

The Raleigh Bicycle Company first introduced Bowden cables as bicycle brake actuators in the later part of the nineteenth century. They were then adopted by the fledgling aircraft industry of the early twentieth century for the control of aircraft flight surfaces. At the end of World War II, many former aircraft designers, in particular engineers at Northrop-Grumman, were set to the task of designing better prostheses for returning U.S. veterans. One of their more important contributions was the use of Bowden cables to operate upper-limb prostheses.

The typical prosthesis control system consists of a Bowden cable with appropriate terminal fittings. The terminal fittings are used to anchor one end of the cable to a body-harness, and the other end to the prosthetic component to be controlled. Between the two end points, the cable
crosses the prosthetic joint to be controlled and the physiological joint that is moved to actuate this prosthetic joint. The housing through which the cable slides acts as a guide or channel for the transmission of force by the cable. Retainers on the housing fasten it to the prosthesis, and serve as reaction points in the transmission of force by the cable.

The basic configuration of Bowden cables in prostheses has changed little over the intervening years and is still in use today. In fact, if prehensile function is the primary goal of the prosthetic fitting, the device of choice for most persons with amputations is a body-powered, Bowden-cable-operated prosthesis with a split hook-shaped terminal device. This is in spite of all the technological advances in electronics, computers, and dc motor technology that have occurred since the end of World War II.

*Low technology does not imply wrong or bad technology.* In fact, ease of maintenance, ease of repair in the event of failures in the field (one can use a piece of heavy cord to get by if the control cable should break), and the intuitive understanding of pulling on one end of the cable to effect motion at the other are probably major reasons for the success of Bowden cables in prosthetics. This is in addition to the ability of users to sense prosthesis state by the pull or feel of the control cable and harness on their skin.

For transradial (below-elbow) prostheses, only one Bowden cable is needed to open and close the terminal device (Fig. 32.5). In transhumeral (above-elbow) prostheses, two Bowden cables are needed, one to lock and unlock the elbow and another to flex and extend the elbow when the elbow

**FIGURE 32.5** Photograph of a person with bilateral transradial (below-the-elbow) amputations using body-powered prostheses to perform standard object manipulation tasks during an occupational therapy session. Tasks of this nature usually take the form of pick and place trials. These trials require the subject to move objects of various sizes and shapes as quickly as possible between two different locations. This teaches the subject how to coordinate the prosthesis control while moving the terminal device. In another set of trials the subject moves objects of various sizes and shapes as quickly as possible from one hand to the other, for the purpose of teaching bimanual dexterity. Such tasks are used for training and evaluation purposes.
is unlocked or to open and close the terminal device when the elbow is locked (Fig. 32.6). Both the transradial and transhumeral prostheses use a harness worn about the shoulders to suspend the prosthesis and for attachment of the control cable.

A combination of glenohumeral flexion and shoulder abduction is the primary body-control motion used to affect terminal device opening and closing or elbow flexion and extension. Typically, a total excursion of 10 cm (4 in) and upward of 222 N (50 lbf) of force are possible using these motions. Elbow lock control is affected by a complex shoulder motion which involves downward rotation of the scapula combined with simultaneous abduction and slight extension of the shoulder joint. Figure 32.7 shows the

<table>
<thead>
<tr>
<th>Control source</th>
<th>Force available</th>
<th>Excursion available</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>N</td>
<td>lbf</td>
</tr>
<tr>
<td>Arm flexion</td>
<td>280</td>
<td>63</td>
</tr>
<tr>
<td>Shoulder flexion</td>
<td>271</td>
<td>61</td>
</tr>
<tr>
<td>Arm extension</td>
<td>247</td>
<td>56</td>
</tr>
</tbody>
</table>

Source: From Taylor (1954).
basic harnessing of these cables for each case. There are many different variations on these basic harnessing schemes depending on specific needs of the prosthesis user. These variations and their application are described in Fryer (1992). Fryer and Michael (1992) provide an excellent overview of available body-powered components.

Typical forces and excursions for different body-powered control sources for an average adult male with an arm amputation can be found in Table 32.1. As can be seen, the average person has fairly high magnitude sources of force and excursion. By varying the locations of the reaction points of the Bowden cable or through the use of a pulley mechanism arranged to amplify force, or excursion, force can be interchanged with excursion or vice versa. However, the total power (force times excursion) remains constant.

In prosthetics, stainless steel Bowden cables come in three sizes: Light Duty, Standard Duty, and Heavy Duty. The basic configuration is a Standard size multistranded steel cable in a Standard size housing. A heavy-duty user will use a Heavy-Duty multistranded steel cable in a Heavy-Duty housing. Friction is the primary cause of transmission losses in a Bowden cable. If friction is an issue then a Heavy duty housing with a Teflon liner and a Standard size cable can be used.

Composed of ultra-high-molecular-weight polyethylene fibers, Spectra cable is another option that is available to body-powered prosthesis designers and users. Spectra cable is now widely used at almost all major upper-limb fitting centers and could be considered the standard for contemporary fittings. Spectra’s is used extensively as fishing line for deep-sea sport fishing because one of its properties is that it does not stretch. In addition, Spectra is very lightweight, quiet, strong, and it has a low coefficient of friction, making it an ideal alternative for numerous prosthetic applications, especially for children. Spectra is almost certainly always used when friction losses need to be avoided.

In Europe it is common to use a Perlon cable in a standard housing for light to medium users. While Spectra, like Perlon, can be used with standard cable housings, lower friction can be achieved by using Spectra or Perlon with a Heavy Duty housing with a Teflon insert. Note, the Heavy Duty housing is not needed for strength but for its wider inside diameter to accommodate the liner. This way one has plastic on plastic, reducing wear and friction and increasing lifetime. It should be noted that standard swage fittings will not work with Spectra and that tie-off fittings are required. Spectra is a registered trademark of Allied-Signal, Inc.

Tip. Bowden housings are essentially tightly wound steel spirals (tightly wound springs) and as such can be screwed into an appropriately sized hole. For a Standard housing use a 1/8-in drill and a No. 8–32 Tap (note that this is a nonstandard Drill and Tap combination).

Body-powered prehension devices can be either voluntary opening or voluntary closing. A voluntary opening device requires a closing force, usually exerted by rubber bands, to be overcome before it will open (default to close position). The maximum pinch force in a voluntary opening device is limited to the closing force exerted by the rubber bands. Typically, a standard prosthesis rubber band results in 7 to 9 Newtons (~1 1/2 to 2 lbf) of prehension force and a healthy consistent user will have anywhere from 3 to 8 rubber bands depending on strength, amputation level, and personal preference.

32.3.5 Voluntary Opening Versus Voluntary Closing Devices

FIGURE 32.7 Schematics showing the harnessing and control motions used in both transradial and transhumeral body-powered prostheses. (a) Glenohumeral flexion—forward motion of the upper arm about the shoulder. This schematic shows the harnessing for a transradial prosthesis, (b) Glenohumeral flexion for a person harnessed for a transhumeral prosthesis, (c) Biscapular abduction (rounding of the shoulders) used by person with either transradial or transhumeral amputations. A combination of glenohumeral flexion and biscapular abduction is the most common mode of body-powered prosthesis control, (d) Shoulder depression followed by glenohumeral extension is the control motion used by persons with transhumeral amputations to actuate the elbow lock.
Voluntary closing devices require an opening force to be overcome before the terminal device will close (default to open position). Pinch force in a voluntary closing terminal device is directly proportional to the force applied to the device through the control cable. The voluntary closing principle is the closest to natural hand prehension (Fletcher, 1954). However, tension must be maintained in the control cable to hold a constant gripping force or position, just as must be applied by the human hand to maintain grasp. The natural neuromuscular mechanism has the ability to maintain grasp over long periods of time without undue strain. Automatic or manual prehension locking mechanisms have been used for this function in voluntary closing devices. A voluntary opening device does not have this problem as long as the object held can withstand the maximum closing force.

The Therapeutic Recreation Systems (TRS, Boulder, Colo.) range of terminal devices and the Army Prosthetic Research Laboratory (APRL) hook, which was used predominantly by muscle tunnel cineplasty amputees, are examples of successful voluntary closing devices that are available. The APRL hook has automatic locking, whereas the TRS devices do not.

Operating requirements (force and displacement) for a number of standard body-powered prosthetic components are given in Table 32.2. Neither the Army Prosthetics Research Laboratory Hook nor the Sierra Two-Load Hook are used much today. It is most common to find a voluntary opening split hook on current body-powered systems. However, if one were to compare new contemporary fittings with refitting of existing prostheses, the percentage of voluntary opening split hooks is declining among new contemporary fittings in the United States. At the transradial level, the TRS devices are becoming the prehensor of choice with sales of TRS devices far exceeding sales of APRL hooks and hands. Properly harnessed, the TRS devices are nearly closed in the relaxed, or neutral, position.

### 32.3.6 Electric Power Sources (Batteries)

Battery technology, specifically rechargeable battery technology, is vital to portable electronic equipment and is driven by the billions of dollars spent by the laptop computer and cellular phone industries. The field of prosthetics and orthotics (P&O) sits on the sidelines and picks up anything that looks like it could be of use. In an electrically powered prosthesis, the main current draw comes from the dc motor(s) used to actuate the device. In a dc motor, the output torque is directly proportional to the amount of current drawn. Motor use in prostheses is not continuous but is intermittent. Consequently, it is important not only to know how much energy a battery can provide but also how fast the battery can provide it.

The maximum amount of current drawn by a dc motor is the **stall current**. This is the current drawn by the motor when it is fully “on” but unable to move, such as occurs when a hand has grasped an object and the fingers can close no further. This is also the point of maximum torque.

---

**Table 32.2** Force and Excursions Needed to Actuate a Number of Standard Body-Powered Prosthetic Components

<table>
<thead>
<tr>
<th>Component/operation</th>
<th>Force needed</th>
<th>Excursion needed</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow Flexion (No Load on Hook)</td>
<td>40 N, 9 lbf</td>
<td>51 mm, 2&quot;</td>
</tr>
<tr>
<td>Prehension, APRL Voluntary Closing Hook</td>
<td>40–155 N, 9–35 lbf</td>
<td>38 mm, 1.5&quot;</td>
</tr>
<tr>
<td>Prehension, Sierra Two-load* Voluntary Opening Hook</td>
<td>44.4, 89 N, 10, 20 lbf</td>
<td>38 mm, 1.5&quot;</td>
</tr>
<tr>
<td>Elbow Lock Actuation</td>
<td>9–18 N, 2–4 lbf</td>
<td>15 mm, 0.6&quot;</td>
</tr>
</tbody>
</table>

*The Sierra Two-Load Hook, formerly the Northrop Two-Load Hook, was a voluntary opening hook with two closing force settings. A manually controlled switch on the side of the hook moved the hook between force settings.

output of the motor and is known as the \textit{stall torque}. Running a dc motor in stall for extended periods of time may damage the motor. However, the stall current is the upper-limit on the amount of current required by a dc motor driven mechanism. As such, the stall current determines the size of the MOSFET’s (field-effect transistor) and/or H-bridges used to deliver current to the motor.

Batteries are a chemical means of producing electricity, in which the choice of electrode materials is dictated by their relative location in the electrochemical series. Compounds that are at the extremes of the series are desirable. Many battery chemistries exist. Batteries are usually packages of a number of cells stacked in series to achieve a desired voltage. The voltage of the individual battery cells depends on the chemistry of the cell. The nonrechargeable off-the-shelf batteries that one buys in the store are referred to as \textit{alkaline batteries}. In P&O we are most interested in rechargeable batteries with solid chemistries, i.e., nonliquid batteries.

Size, weight, capacity, and power density are the primary selection considerations for batteries in externally powered prosthetic design applications. The most popular types of rechargeable batteries in use in prosthetics today are nickel-cadmium (NiCd), nickel-metal-hydride (NiMH), and lithium-ion (Li-ion). Li-ion is fast becoming the chemistry of choice because of its high capacity-to-size (weight) ratio and low self-discharge characteristic.

The amount of time it takes to discharge a battery depends upon the battery capacity \( C \) expressed in milliamp hours (mAh) (more is better) and the amount of current drawn by the load. Battery charge and discharge currents are normalized with respect to battery capacity and are expressed in terms of C-rate. C-rate = \( C/1 \) hour, e.g., to a 150-mAh battery has a C-Rate of 150 mA. The current corresponding to 1C is 150 mA and 0.1C, 15 mA. For a given cell type, the behavior of cells with varying capacity is similar at the same C-rate. More in depth information about batteries can be found in the \textit{Handbook of Batteries} (Linden, 1995).

Battery capacity is the important measure of performance in the cell phone and laptop industries because once the device is switched “on,” the load current remains fairly constant. The \textit{discharge rate} is an important measure for prosthetics applications because it is a measure of the rate at which energy can be delivered to the load. It is the maximum allowable load or discharge current, expressed in units of C-rate. The higher the discharge rate, the faster a battery can meet the demand for energy.

The \textit{self-discharge rate} is the rate at which a battery discharges with no load. Li-ion batteries are a factor of two better than NiCd or NiMH batteries. The number of charge and discharge cycles is the average number of times a battery can be discharged and then recharged and is a measure of the service life. For prosthetics applications, a battery ought to provide at least a day of use before needing to be recharged; thus the average life expectancy for a rechargeable battery in prosthetics use should be about 3 years. Table 32.3 tabulates these quantities for the chemistries of interest.

The problem of \textit{memory} that one hears about in association with NiCd batteries is relatively rare. It can occur during cyclic discharging to a definite fixed level and subsequent recharging. Upon discharging, the cell potential drops several tenths of a volt below normal and remains there for the rest of the discharge. The total ampere-hour capacity of the cell is not significantly affected. Memory usually disappears if the cell is almost fully discharged (\textit{deep discharge}) and then recharged a couple of times. In practice, memory is often not a problem because NiCd battery packs are rarely discharged to the same level before recharging. Environmental concerns exist regarding the proper disposal of NiCd batteries because of the hazardous metal content. NiMH and Li-ion batteries do not contain significant amounts of pollutants, but nevertheless, some caution should be used in their disposal. Discharge profiles for these three popular types of batteries are shown in Fig. 32.8. Note that NiCd batteries have a greater discharge rate than LiM batteries (making them suitable for prosthetic applications) however their energy density is lower.

\textit{Slow charging} (charging times greater than 12 hours) is straightforward for all battery chemistries and can be accomplished using a current source. Charge termination is not critical, but a timer is sometimes used to end slow charging of NiMH batteries. Li-ion slow-chargers should have a voltage limit to terminate charging of Li-ion batteries. \textit{Trickle charging} is the charging current a cell can accept continually without affecting service life. A safe trickle charge for NiMH batteries is typically 0.03C. \textit{Fast charging} (charge time of less than 3 hours) is more involved. Fast-charging circuits must be tailored to the battery chemistry and provide both reliable charging and charge termination.
Overcharging can damage the battery by causing overheating and catastrophic outgassing of the electrolyte. The gas released from any outgassing may be dangerous and corrosive. In some instances overcharging may cause the battery to explode (Kester and Buxton, 1998).

Standard operating voltages used in prosthetics are 4.8, 6, 9, and 12 V depending on the manufacturer and component to be driven. All Otto Bock System 2000 children’s hands (Fig. 32.14)
Otto Bock adult hands (Fig. 32.12), Steeper hands/prehensors (Fig. 32.20c) (RSLSteeper, Rochester, U.K.), VASI hands and elbows [Variety Ability System, Inc. (VASI), Toronto, Canada] and Hosmer NYU elbow (Hosmer-Dorrance Corp., Campbell, Calif.) are specified to run off 6 V but can sometimes be run at higher voltages to boost performance. The Hosmer NU-VA Synergetic Prehensor (Fig. 32.20b) and Motion Control ProControl System (Motion Control, Inc., Salt Lake City, Utah) use 9 V. The Motion Control Utah Arm and LTI Boston Elbow [Liberating Technology, Inc. (LTI), Holliston, Mass.] run off 12 V, while Motion Control’s electric hand is specified to run on any voltage between 6 and 18 V. In general, adult-size electric hands use 6 V while adult-size electric elbows use 12 V. Elbows need the extra voltage so that they can have a useful live-lift capacity (torque to actively lift a load). Typical currents are 1 to 3 A (~12 to 36 W) for hand mechanisms and 9 to 12 A (~120 W) for some of the brushless dc motors used in elbow prostheses.

The preponderance of 6-V electric hands is due to the dominance and history of Otto Bock in the field of prosthetics. Otto Bock was one of the first companies to produce an electric hand with its associated electronics and power source in the early 1970s. Their standard battery was the 757B8 five-cell NiCd battery pack, where each cell has a voltage of 1.2 V. Otto Bock’s current standard battery is now a Li-ion battery. This is a custom-designed battery pack and battery holder. The battery holder is designed to be laminated into the prosthetic socket so that batteries can be easily interchanged should one discharge completely during use. Even if the battery chemistry has changed, this battery holder’s form has been around for many years. The disadvantage of a custom form is that if the battery dies in the field and the user does not have any spares then they are stuck without a working component. Liberating Technology, Inc. and Motion Control elbows use custom battery packs, whereas the Hosmer-Dorrance elbows use packs made up of off-the-shelf AA rechargeable batteries (Fig. 32.9).

Our laboratory’s preference is to use rechargeable batteries in a 9-V transistor battery form. This enables the prosthesis to be recharged overnight, while a standard 9 V transistor battery form allows commercially available off-the-shelf 9-V alkaline batteries to be easily purchased and used should the rechargeable battery run out of charge unexpectedly. Motion Control sells an Otto Bock battery case that has been modified to accommodate a standard 9-V transistor battery.

32.3.7 DC Electric Motors

By far the most common actuator for electrically powered prostheses is the permanent magnet dc electric motor with some form of transmission (Fig. 32.10). While there is much research into other electrically powered actuator technologies, such as shape memory alloys and electroactive polymers, none is to the point where it can compete against the dc electric motor. A review of the available and developing actuator technologies with their associated advantages and disadvantages as well as their
power and force densities can be found in Hannaford and Winters (1990) and Hollerbach et al. (1991). In both these reviews, biological muscle is used as the benchmark for comparison of these technologies.

For electrically powered prosthetic hand mechanisms, a coreless, or ironless, dc motor with a fitted gear head transmission is the actuator of choice. For elbows, coreless dc motors, or brushless dc motors and a transmission are used. Brushless dc motors are becoming more common now that both the motor and its electronics are small enough to be accommodated in a prosthetic elbow (they are still too large for hand mechanisms, but this is likely to change). Although brushless motors require much more complicated control electronics, their use is justified because they have substantially higher performance than their coreless counterparts. In addition, recent advances in surface mounted integrated circuit (IC) technology greatly facilitate the design of controllers for these motors. A broad range of driver and controller ICs are available in surface-mount forms from companies like Texas Instruments, International Rectifier, ST Electronics, Vishay-Siliconix, Zetex, among others, and application notes explaining the use of these chips are readily available on company Web pages.

### 32.3.8 Sizing a DC Motor for Maximum Performance

In general, because there is no actuator technology that can match muscle tissue’s speed and force capabilities for a given weight and volume, one chooses the largest motor-gearhead combination that will fit within the physical constraints of the mechanism being designed. When one discusses maximizing performance, one is talking about maximizing output speed and output torque to attain physiological speeds and torques. Almost without exception, powered prosthetics components use MicroMo-Faulhaber (Faulhaber Group, Clearwater, Fla.) motors. This company manufacturers coreless motors, brushless motors, and their associated gearheads. Their motors range in diameter from 6 up to 35 mm.

The key issue to keep in mind when choosing a motor is that one trades no-load speed for stall torque and vice versa (Fig. 32.11). A prosthetic hand usually operates at high speed and low torque during hand closing and opening and then at high torque and low speed while gripping an object. To meet both these requirements, assuming an automatic transmission is not used, a single motor must be sized to provide both the requisite stall torque as well as the requisite no-load speed. Thus, the motor...
must be capable of providing the mechanical power output given by the product of the no-load speed times the stall torque.

Choice of gear ratio depends on the no-load speed and stall torque of the motor selected, as well as on the desired output speed and torque of the mechanism. Remember speed times torque into and out of the gear train is constant (assuming an ideal gear train). Usually, when using a dc motor, speed

**FIGURE 32.11** (a) Typical speed-torque and current-torque relationships for a dc motor. The key issue is that stall torque times no-load speed is constant (stall torque × no-load speed = const.) regardless of the ratio of the gear train driven by the motor. The main points of interest on this curve are no-load speed (output speed of the motor when running without load), the stall torque (torque generated by the motor when fully "ON" but unable to move), and the area under the curve (which is the total power available to the motor). Typical units are included in brackets—imperial equivalents are in the square brackets (careful use of the correct conversion factors is required when working with imperial units). (b) Adding a gear head to the motor output changes the no-load speed and stall torque of the drive train, but the area under the new speed-torque relationships remains the same as for the motor without the gearhead (assuming a lossless gearhead, i.e., 100 percent efficient). One trades no-load speed for stall torque and vice versa, but the total available power remains the same.
is traded for torque. One must also allow for the overall efficiency of the gearhead and the maximum allowable intermittent output torque for a given gearhead. The MicroMo catalog (www.micromo.com) provides worked examples of how to choose the correct motor and gearhead for a particular output speed and force requirement (Faulhaber MicroMo Application Notes).

The maximum allowable intermittent output torque is a function of the tooth strength (this is a material property) of the gears and, as such, is an upper limit on the torque a particular gear can handle. Thus, although a particular gear ratio might suggest that the output torque is attainable, the gears in the drive train may not be physically able to handle the loads placed on them. Planetary gear trains can usually handle greater torques than simple spur gear trains, but usually at the expense of efficiency. A further constraint is to ensure that the motor-gearhead maximum axial load will not be exceeded. This is usually not an issue when using spur gears on the output. However, if a worm gear, lead screw, or some other rotation-to-linear conversion mechanism is used then this parameter must be taken into consideration and a thrust bearing used to protect the gearhead if necessary. A good reference is the Handbook of Small Electric Motors (Yeadon and Yeadon, 2001).

Often, for a given size of motor, a manufacturer will offer motors with different magnet materials. To maximize the performance (force-speed output) of a particular size of motor, choose the “rare earth” version (neodymium-iron/neodymium-iron-boron magnets). (Note: rare-earth magnet motors tend to pull higher currents in stall.) In addition, when specifying the nominal drive voltage of the motor, consider deliberately overvoltageing the motor to boost overall performance. This can be done because prosthetic use is intermittent in nature. A nominal 6-V motor can be run at 9 V yielding a ~50 percent increase in output speed and torque. Care must be taken, however, because the stall current will also increase. Prolonged exposure to increased stall current can damage the motor unless the rise in current as the motor stalls is detected and the supply voltage cut off, or the current limited. A final word of caution, ultimate drive system torque performance is limited by the amount of current that the battery can provide. If the battery cannot provide the stall current needed then the drive system will not reach the torque levels suggested by your calculations.

Current flowing in the motor can be monitored using a number of different ICs or power FETs available from any number of suppliers. These devices include current shunt ICs (Maxim, Texas Instruments Zetex), sense FETs (Fairchild Semiconductor), HEX sense power FETs (International Rectifier), and fault-protected switches (fancy MOSFETs, Micrel) to name but a few. Current-shunt ICs are common because they find use in dc-to-dc converters, consequently, a large number of different manufacturers make them in many different forms and flavors (see Fig. 32.17).

Figure 32.17 shows an implementation of high-side current monitoring on a n-channel H-bridge. (Note: this circuit is not limited to use with just n-channel bridges.) In this circuit a shunt resistor measures current flowing to the bridge, and an amplifier scales the voltage from the shunt resistor to logic levels, where a 5-V signal corresponds to the maximum current anticipated. This voltage level is compared against some threshold level to ensure that too much current is not being drawn. If it is, the comparator output goes high or low depending on your logic of choice (logic-low will fail-safe if the power goes out) and another supervisory circuit (microprocessor, etc.) takes action to switch off the motor. This same circuit can be implemented as a low-side current monitor just by placing it between ground and the bridge. Since the bridge is under your control, you should know the motor’s direction. If current sensing for each motor direction is required, then this circuit needs to be duplicated and a sense resistor placed in each side of the bridge at the location of the switches shown in the pn complementary H-bridge.

An implementation of limit switch protection of the motor is shown in Fig. 32.17 on a pn complimentary H-bridge. In this case electromechanical switches are physically located at the ends of the range-of-motion of the component (full flexion or full extension of an elbow, for example). When the elbow extends fully it physically hits/bumps/actuates the switch, causing it to open. This has the effect of “dropping out” the ground line for the extension direction motor supply, causing the motor to stop extending. Further extension drive signals will not extend the component. A drive signal of the reverse polarity travels through the other side of the H-bridge enabling the motor to run in the opposite direction (flex in this example). Again this example is not limited to pn bridges and could also be implemented on the high side.
While the output power of a motor may remain constant, how that power is used, i.e., whether all the power is used to generate high force at low speed or high speed at low force, is up to the designer. Since there is no one motor small enough to be placed in an artificial hand that can meet both the speed and force requirements (even with overvoltage), other techniques must be employed to increase hand mechanism speed of opening and closing and grip force. Two techniques are currently employed in commercially available prosthetic prehensors to increase the performance of prosthetic hands to levels approaching those of the physiological hand.

The first technique is to use a single large motor and an automatic transmission. In this case, the motor opens and closes the hand in a high-speed, low force gear ratio. When the hand closes against an object, the rise in force (as detected by an internal spring) automatically triggers the transmission to switch into a high-force, low-speed mode that allows the hand mechanism to build up prehension/pinch force. The main disadvantage of these mechanisms is associated with tightly grasped objects. In this instance, during opening, a perceptible amount of time appears to elapse before the hand visibly opens. This is so because time must be allowed for the force built up during the grasping cycle to drop below the force threshold where the hand automatically switches from high force, low speed to low force, high speed. Thus, although the hand is opening, the gear ratio is such that little motion occurs until the transmission switches into the high-speed mode. To the user, this time spent “unwinding the hand” appears as if nothing is happening in response to the open command. A further disadvantage of automatic transmissions is that they are mechanically complex and are often beyond the means of the individual to design in a form compact enough for use in hand prostheses. The Otto Bock range of System Electric Hands (Fig. 32.12) for adults has used and refined its automatic

![Otto Bock System Electrohand](image)

**FIGURE 32.12** Otto Bock System Electrohand (Otto Bock Orthopedic Industry, Inc., Duderstadt, Germany). The hand consists of a mechanism over which a liner is placed. A cosmetic glove is then pulled over the liner. (Note: Otto Bock does not provide silicone gloves.) Liner creases have been placed in the finger thumb web space in an effort to reduce its elastic resistance. Also shown is a pair of tweezers that are provided with each hand to make tip prehension possible. Prosthetic hands have very poor tip prehension without the aid of some sort of extra tool such as these tweezers.
transmission for many years, to the point where their design is a marvel of compactness and robustness: Why reinvent the wheel? The Motion Control hand, which owes much of its mechanical design to the Otto Bock hands, also uses an automatic transmission.

The second technique is to use multiple smaller motors configured for what Childress (1973) called synergetic prehension. In synergetic prehension, one motor is geared for high speed and low force, and another is geared for high force at low speed. A simple synergetic prehensor consists of two motors that open and close a split hook (Fig. 32.13a). One motor gives one tine of the hook high speed and excursion but little force (fast side), the other motor gives the other tine of the hook high force but little speed and excursion (force side). In this way the motors work in synergy to boost overall prehension performance (Fig. 32.13b).

![Figure 32.13](image-url)

**FIGURE 32.13** (a) Schematic of the two-motor concept of synergetic prehension. Motor 1 is geared to provide high speed at low force, whereas motor 2 is geared to provide high force at low speed. The two motors operate in synergetic prehension to provide a system with high speed and high force. (b) The effective speed torque relationship (C) of two drive systems having characteristics A and B operating in synergy. A is the characteristic for a high-speed, low-torque motor, whereas B is the characteristic for a low-speed, high-torque motor. D is the required speed-torque relationship for a single dc motor and gearhead to achieve the same no-load speed and stall torque as the synergetic system.
32.3.9 Synergetic Prehension

The principle of synergetic prehension stems from Childress’ (1973) observation that the act of grasping an object usually requires little real work. When an object is grasped, a force is exerted with very little excursion, whereas excursion of the grasping fingers when approaching the object usually occurs in space and requires very little force. In both cases, the work involved is minimal. The exception is grasping a compliant object, where both force and excursion are required (e.g., squeezing a lemon). This condition, however, is not the usual case in prosthetics. Synergetic prehension can be readily implemented using multiple motors that operate together, or in synergy. Each synergetic motor pair controls a single degree of freedom and has its own synergetic motor controller. The combination of the synergetic controller and synergetic motor pair allows it to be treated like a single motor. Placing a voltage across the input lines of the synergetic controller will drive the motors in the synergetic motor pair. The polarity of this voltage determines the direction the motors in the pair will rotate. Drive signals in the form of either pulse-width-modulated (PWM) signals for proportional speed and force control or on/off signals for switch or single speed control can be used.

The Hosmer-Dorrance NU-VA Synergetic Prehensor was developed using this theory and achieved angular velocities in excess of 3 rad/s and prehension forces greater than 20 lbf (89 N) using a small 9-V transistor battery (Fig. 32.20b). The Otto Bock System 2000 children’s hands (except for the smallest size) (Fig. 32.14) and the RSL Steeper Powered Gripper (RSL Steeper, U.K.), also use the principle of synergetic prehension (Fig. 32.20c). An advantage of synergetic systems is that they can potentially develop high force and high speed at low power. The main disadvantage associated with synergetic systems is the use of multiple motors. More motors mean more parts that can fail. In addition, the motor and its gearhead tend to be some of the most expensive components in a hand mechanism.

Not all commercially available hands use automatic transmissions or multiple motors operating in synergy. The Variety, Ability System Inc. (VASI, Canada) range of children’s hands, Centri AB’s

FIGURE 32.14 Otto Bock System 2000 children’s hand (Otto Bock Orthopedic Industry, Inc., Duderstadt, Germany). This mechanism uses two gear motors attached end to end that operate using the principle of synergetic prehension. Notice that this hand does not use a liner. The cosmetic glove is pulled directly over the mechanism. This is so because for a child’s hand, the forces are not as high as for an adult mechanism and the tips will not punch through the glove at the lower grip force.
(Sweden) hands, and RSLSteeper’s Adult Electric Hands (RSLSteeper, U.K.) are all examples of single-motor designs with a fixed gear ratio. Typical performance for these devices is correspondingly lower [35 N (8 lbf) pinch, 10 cm/s (4 in/s) speed] than typical performance for those prehensors that utilize synergy or automatic transmissions [80 N (18 lbf) pinch, 11 cm/s (4.3 in/s) speed].

All hands need some form of backlock mechanism to prevent the fingers from being forced open or backdriven once power is removed. This is so that a hand mechanism will hold its position, or applied force, once the control signal is removed. While this is nonphysiological, (i.e., muscles must keep contracting to maintain a position or an applied force), this preserves power. Synergetic mechanisms need a backlock mechanism to prevent the force side from backdriving the speed side, in addition to maintaining applied force in the absence of power.

The RSLSteeper Electric Hands use a worm gear for this purpose. A worm gear is inherently nonbackdrivable. The VAS hands, the NU-VA Synergetic prehensor, and the LTI Boston Elbow all use variations on a roller clutch (backlock) first used in ViennaTone hands of the 1960s (Fig. 32.15). Locking occurs when a cam inside the mechanism wedges a roller(s) between itself and the outside cylindrical surface in response to an external torque. Driving the input shaft from the motor side causes the cam to move away from the roller(s), allowing the roller(s) to move away from the outside surface, thus freeing up the mechanism. These roller backlocks can resist external torques in either direction or only one direction depending on the number and position of the roller(s) and shape of the cam. Roller backlocks should be placed as high up in the gear train (close to the motor) as possible to minimize backlash (play) and to minimize the force exerted on the cylindrical surface as well as the force needed to unlock the mechanism. Machinery’s Handbook (Oberg et al., 2000) is a good reference on different types of mechanism and all issues pertaining to the design and machining of mechanical components, including material properties, tooling, machining, manufacturing, gearing, threading, fasteners, etc.

Backlocks are also found in some prosthetic elbows (Boston Elbow III, LTI; Otto Bock Body-Powered Elbow) as a means of holding elbow position in the absence of a drive torque. This ability allows users to park their prosthetic arm at any desired position and then remove (decouple) themselves from the arm’s control.

The Otto Bock Body-Powered Elbow mechanism uses a springlike clamp that locks down on a shaft when twisted in one direction but opens to release a shaft when twisted in the other. This is a unidirectional mechanism whose main advantage is simplicity. A similar mechanism can be found in some children’s bicycles, where the child backpedals to brake the bike. A disadvantage of this type of mechanism is that over time the lock can slip because of wear and tear on the spring on the elbow shaft.

### 32.3.10 Electrical Power Distribution

The torque output of a dc electric motor is linearly related to the amount of current the motor can handle. The limiting factor is usually heat. Current flow in the windings of the motor generates heat. If the heat in a motor can be dissipated fast enough, then the motor is able to handle more current. Thus heat sinks can be used to dissipate heat generated and boost motor performance. The same issue applies to the electronic components used to switch the direction of current used to control the motor. In order to drive the motor, an H-bridge, generally made up of four power MOSFETs, is usually employed to direct current to the motor under the direction of a controller. As a rule, the higher the current, the larger the MOSFETs in the H-bridge must be to handle the current and to dissipate the heat generated, although heat sinks can once again be employed to enable smaller MOSFETs handle larger currents.

The important issues from a prosthetics standpoint for the selection and/or design of bridge circuits is that they be as small as possible and consume as little power as possible. In general, one wants as much of the battery’s power as possible to be directed to drive the mechanism motors rather than expending it running the prosthesis/bridge controller. An empirical rule of thumb based on experience in our laboratory is that no more than 5 percent of the available energy should be used to power the controller circuitry.
MOSFET H-bridges come in two basic configurations: n-channel half-bridge and p- and n-channel (complementary) half-bridge (Fig. 32.16). Simplicity of the MOSFET's gate drive is the main advantage of the p- and n-complementary half-bridge. When an n-channel MOSFET is used for the high-side (or upper) switch, the gate drive signal requires level shifting, resulting in increased complexity and cost. However, most power IC processes are optimized for n-channel devices. Also, n-channel power MOSFETs (n-FETs) are more efficient than p-channel power MOSFETs (p-FETs) in terms of die size for a given current and voltage. For a given breakdown voltage rating, a p-FET can be 2.5 to 4 times the area of a comparable n-channel device.

The H-bridge and/or MOSFETs are some of the larger components required in the electronics used to control a prosthetic mechanism (Fig. 32.17). Surface-mount technology is employed in
FIGURE 32.16 There are two types of power MOSFETs: $n$ (top left) and $p$ (top right). When used to drive a motor or a speaker, these MOSFETs are usually configured as either an $n$-channel half-bridge (bottom left) or a complementary ($pn$) half-bridge (lower right). One talks about half-bridges because when driving multiphase motors a half-bridge is needed to drive each phase. Brushless motors need three half-bridges, whereas dc motors need two. Often a logic circuit with logic circuit voltage levels (5 V) is used to drive the bridge FETs. To ensure that the FET turns on fully, it is common to use a pull-up resistor, as shown in the bottom diagram, to improve FET performance. Also shown are the associated pinouts for a standard TO-220 package. Note: FETs do not normally have logic-level gate drive signals, and consequently, this must be specified when ordering them.
FIGURE 32.17 Diagram shows a full n-channel and pn (complementary) channel H-bridges with their associated control. It is important to notice that for the n-channel H-bridge diagonally opposite n-FETs are controlled as a pair, whereas for the pn H-Bridge, FETs on the same side of the bridge are controlled as a pair. Also notice how the p-FET is wired as compared with the n-FET in terms of their drain and source connections. The use of the inverter symbol is to highlight that both sides of the bridge are never on at the same time, and in fact, break-before-make switching ought to be used to prevent current transients during the switching process. An implementation of high-side current monitoring is shown on the n-channel H-bridge while an implementation of limit switch protection of the motor is shown on the pn complimentary H-bridge. (See text for further discussion.)
prosthetics control electronics packages to reduce overall circuit size. Surface-mount IC bridges and bridge drivers are available, but there is a problem. In prosthetics the supply voltage is usually 6 to 12 V. In the field of industrial motor control, the standard operating voltage, for which all the commercially available high-current IC bridges and controllers are designed, is anywhere from 12 to 60 V. This means that one must build one’s own bridge using four discrete MOSFETs and some sort of MOSFET driver. While this appears straightforward, H-bridge design can be something of a black art, particularly if two n-channel half-bridges are used, and the resulting bridge can take time to get working properly and can occupy a large amount of printed circuit-board (PCB) space. An alternative is to use standard motor control components and a charge pump to boost the supply voltage to a voltage sufficient to drive both upper and lower MOSFET gates directly. (Vishay-Siliconix, 1992). The use of a charge pump resolves the issue of how to drive the gate of the high side (upper) n-FET above the half-bridge supply to fully turn the device “on.”

Another possibility would be to use a number of low-current (1 A maximum), surface-mount, single-chip H-bridges (e.g., Vishay-Siliconix Si9986 or Zetex) and wire them in parallel. These chips use complementary p- and n- half-bridges in a standard 8-pin SOIC surface-mount package and as such can be run on supply voltages ranging from 4 to 13 V and have a smaller footprint than the standard TO-220 surface-mount package used for some power MOSFETs. Also, using single-chip bridges, rather than building your own, enables a prosthetics designer to benefit from the protection features incorporated into these chips. One such protection feature ensures that both sides of the bridge are never turned on together. Turning on both sides of the bridge at the same time will create a short and burn out the bridge FETs very quickly. Another common cause of FET failure is not turning the FET “on” hard enough (i.e., not turning the FET all the way on or off). If multiple motors are to be used in a hand mechanism these chips are a nice solution since a multiple motor configuration will generally use smaller motors that individually do not draw high currents. A very good introduction to, and general reference on, electronics and electronics design is the Art of Electronics by Horowitz and Hill (1995).

32.3.11 Notes on Pneumatic Power

Historically, the other major source of external power in prosthetics was pneumatic power. Pneumatic actuators were employed in a number of hand and arm designs in the 1970s. Cool and Van Hoorneweder (1971) developed a pneumatically powered hand prosthesis with adaptive, internally powered fingers. This hand could achieve good grasping with low forces because it was able to adapt to the shape of the object grasped. The Edinburgh Arm was a pneumatically powered arm that saw limited clinical use (Simpson, 1973). Unfortunately, this arm was mechanically complex and prone to failure. Kenworthy (1974) also designed a CO₂-powered hand for use with the Edinburgh arm. Otto Bock Healthcare (Duderstadt, Germany) also sold a number of CO₂-powered systems up until the mid-1970s. Disadvantages associated with pneumatics are that a cylinder of gas (carbon dioxide) has to be secreted somewhere either upon the user or in the device and a ready supply of these cylinders must be available. Because the user has to have tubes running from the gas supply to his or her hand, self-containment of the whole prosthesis becomes an issue.

Currently, there are no pneumatically powered prostheses commercially available, but there are a number of groups working on pneumatically powered devices for prosthetic applications. The WILMER group at the Technical University of Delft in the Netherlands is working on a hand mechanism that uses small disposable CO₂ cartridges called “Sparklets” as a power source (Plettenburg, 2002). “Sparklets” are available in retail stores in Europe for making soda water.

What are claimed to be a new class of pneumatic actuator drive a humanoid hand under development at the Karlsruhe Research Center (Schulz et al., 2002). The actuators consist of cavities that change their size when inflated with a pressurized gas or liquid. Forces of up to 10 N can be obtained by driving these flexible pneumatic actuators with air at a pressure of 3 to 5 bar, allowing motion frequencies of up to 10 Hz. The five-fingered hand mimics the functions and resembles the anatomy of the human hand. The thumb consists of two finger segments and an opposition
mechanism; all other fingers consist of three finger segments. All fingers and the wrist can be positioned independently. The compact size of the actuators allows for their full integration into the finger. A safe grasp is accomplished through the flexible self-adaptability of the actuators. The maximized contact surface allows for the use of relatively low pressures.

The BioRobotics Lab in Seattle is experimenting with pneumatic McKibben muscle actuators to mimic the musculature of the natural arm. These actuators, which were first conceived of in the 1950s, consist of an inflatable elastic tube covered with a flexible braided mesh. When pressurized, the elastic tube inside expands but is constrained by the mesh. The flexible mesh shortens or contracts like a muscle due to the expanding tube. It has been found that McKibben muscles can exhibit passive behavior very similar to biological muscle since both have series and parallel elasticity (Klute et al., 1999). From a robotics perspective McKibben muscles can have a high force-to-weight ratio. However this ratio ignores the weight of the compressed air source and associated tubing, which is an issue for a self-contained prosthetic device. In addition McKibben muscles have a shorter range of motion than human muscle and so far have not proven themselves to be reliable long-term mechanisms.

The major attraction of pneumatic systems for prosthetics applications is their inherent compliance, which tends to give these systems a very natural look and feel. While pneumatic systems did find some measure of success in prosthetics, hydraulic systems did not. Hydraulics tended to be messy, with hydraulic units leaking hydraulic fluid. In addition, a fluid reservoir and fluid are required, adding to the total weight of the mechanism.

32.3.12 Note on Hybrid Systems

When body-powered and externally powered systems are linked together, they are called hybrid systems. Hybrid systems are used most frequently with persons who have high-level amputations, that is, amputations above the elbow, or who have bilateral arm amputations. Such systems can provide the user with the high gripping and/or high lifting capacities of powered systems and the fine control of body power. For example, providing a person with bilateral limb loss at the shoulder level with a body-powered limb on one side and with an electrically powered limb on the other side decouples the limbs (Fig. 32.23). The prosthetic fitting shown in Fig. 32.23 highlights a number of important issues.

First, by providing the person with a body-powered limb on one side and an electric-powered limb on the other side, the wearer is able to use the limb that is most appropriate for a specific task. This method also decouples the limbs so that body motions used to operate the body-powered side do not influence the state of the electrically powered limb and vice versa, control of each side is independent of the other.

Second, this fitting demonstrates the highly modular nature of upper-limb prosthetics today—components from many different manufacturers were used in a mix-and-match approach to obtain the most functional set of prostheses for the user.

Third, this fitting highlights the severity of the control interface problem when faced with amputations at this level. Everything is used—conventional harness control and biomechanical switches, using the chin, are extensively used in the control of the body-powered prosthetic components. These same chin motions are used to control externally powered components through the use of electromechanical switches and linear transducers.

Fourth, this fitting demonstrates the high degree of creativity required on the part of the prosthetists in being able to customize and modify prosthetic components meant for other applications and the high degree of customization that is required in general, in the fabrication of these high-level prosthetic fittings. This particular set of arms did not “just happen” but is the result of iterative and incremental improvements that have occurred, as new components and devices have become available, over a period of some ten or more years.

Finally, the use of locking mechanisms to switch control from one component to the next allows for a kind of two-state impedance control. The prosthesis is either in a state of high impedance—rigid or locked, or in a state of low impedance—free or unlocked. The prosthesis, when locked, has very high impedance and becomes a rigid extension of the user. The user can use this rigid extension to
interact with the real world by pulling and pushing on objects. Because the prosthesis is rigid the user “feels” the forces exerted on the prosthesis and has a sense of feedback about what the prosthesis is doing. The user has extended his or her proprioception into the prosthesis, similar to how a person extends their proprioception into a hammer when hammering a nail. This is the embodiment of Simpson’s (1973) extended physiological proprioception (EPP). The other state is when the device is unlocked, now it has very nearly zero impedance. The device in an unlocked state also embodies EPP control. The cable linking the user’s physiological joint to the selected prosthetic joint enables the user to control the position and speed of the prosthetic joint directly through the movement of the physiological joint. Forces exerted on the prosthesis are reflected back through the control cable to the physiological joint. The coupling of the control cable allows the user to extend proprioception into the prosthesis even during movement of the prosthesis.

32.3.13 Notes on Unpowered or Passive Prostheses

There is a class of terminal devices that do not offer prehensile function. Devices in this class, usually hands or adapted tools, are regarded as passive or passively functional prostheses. They have no moving parts and require no cables or batteries for operation. They are typically lightweight and reliable. Generic (standard) passive hand prostheses may consist of a cosmetic outer glove over a soft plastic hand with wire reinforcements in the fingers. Traditionally, cosmetic gloves have been made of polyvinyl chloride (PVC), although silicone is becoming the material of choice (Fig. 32.3). Individualized hands, when expertly done, have a preferable appearance to generic hand replacements. Highly realistic hands, fingers, and finger parts can be custom sculpted and painted to an individual’s size and skin coloring. Such prostheses confer to persons what Beasley has called the highly important function of “social presentation” (Beasley & de Bese, 1990).

Passive work prostheses may be a simple post to provide opposition, or they may incorporate specialized features to aid in certain occupations. A custom-designed system that serves only one function may aid the wearer more than one that is supposed to be multifunctional. In such cases, the prosthetic device is worn on those occasions when it is needed. These devices range from tool adapters to sports mitts.

In the past, a prosthetic fitting was deemed to have been unsuccessful if the patient did not wear an actively controlled prosthesis for prehensile function. In a recent study, Fraser (1998) showed that amputees used their prostheses most frequently for nonprehensile tasks, such as stabilizing, pushing, or pulling an object. In addition, Fraser (1998) showed that amputees with passive or cosmetic prostheses used their devices for nonprehensile tasks on average just as frequently as amputees with active prostheses. These results show that just because a prosthetic device is passive or cosmetic does not imply that it is not functional.

32.4 ANATOMICAL DESIGN CONSIDERATIONS

32.4.1 Prehension or Grasp

Hand function is generally limited to modes of prehension that are used the most often. Numerous studies of how the hands grasp objects have been performed (Schlesinger et al., 1919; Keller et al., 1947; Napier, 1956; Kamakura et al., 1980, to name but a few). Broadly speaking, hand tasks can be subdivided into nonprehensile functions and prehensile functions. Nonprehensile functions of the hand are functions where grasping is not required; for example, pushing an object, holding an object between the body and forearm, flicking, brushing, percussive motions such as playing the piano, etc. Prehensile hand functions are those cases where an object is grasped and held partly or wholly within the hand. The 6 grasping patterns adapted by Keller et al. (1947) from Schlesinger et al. (1919) 12 patterns, are the most widely accepted in the field of prosthetics (Fig. 32.18) and have endured the test of time.
1. Tip prehension
2. Palmar prehension
3. Lateral prehension
4. Hook prehension
5. Spherical prehension
6. Cylindrical prehension

Napier (1956) described tip prehension, palmar prehension, and lateral prehension as precision grips and spherical and cylindrical prehension as power grasp, whereas hook prehension falls outside both these categories. Precision grips primarily involve the thumb working in opposition with the index and middle fingers. Tip prehension, or fingernail pinch, is used mainly to grasp small objects. In lateral prehension, the thumb holds an object against the side of the index finger, as is the case when using a key. In palmar prehension (sometimes referred to as tridigital pinch or three-jaw chuck), the thumb opposes either a single finger or two or more fingers.

Power grasps use all the fingers of the hand to provide an encompassing grasp that firmly stabilizes the object being held. Hook prehension is achieved by flexing the fingers into a hook; the thumb is either alongside the index finger or opposes the index and middle fingers to lock the object held. Carrying a brief case is a good illustration of this kind of prehension. Keller et al. found that palmar prehension or tri-digital pinch was the most frequently used prehensile pattern for static grasping whereas lateral prehension is used most often for dynamic grasping.

The finding by Keller et al. (1947) that palmar prehension was the most frequently used pattern and reduction of most prosthetic terminal devices to a single DOF has meant that most prosthetic hands incorporate palmar prehension as the dominant grasp pattern. The persistence of this pattern, combined with a wide width of opening in prosthetic hand designs and its general acceptance over the years tends to support this compromise (Heckathorne, 1992).

A study done at the University of California at Los Angeles (UCLA) (Taylor, 1954) on human prehension force indicated that adult males could produce maximum mean forces of 95.6 N (21.5 lbf) of palmar prehension, 103 N (23.2 lbf) of lateral prehension, and 400 N (90 lbf) of cylindrical grasp. In light of another (unpublished) UCLA study that showed that forces up to 68 N (15 lbf) were needed for carrying out activities of daily living, Peizer et al. (1969) proposed that 68 N (15 lbf) be a minimum standard for the maximum prehension force for electric prehensors (Heckathorne, 1992).

32.4.2 Dominant and Nondominant Hands

The issue of hand dominance, whether one is right or left-handed, must also be considered. People use their dominant hand differently from their nondominant hand. The role of the nondominant hand is to hold things while the dominant hand is working or waiting to work on them. A unilateral amputee will always use his or her prosthesis in a nondominant role even if the prosthesis is a replacement for what was once the amputee’s dominant hand. The unilateral amputee will also tend to pick things up with his or her dominant hand and then place them into the nondominant hand. Consequently, van Lunteren et al. (1983) suggested that there is a difference in the type of grasps that ought to be incorporated in devices for unilateral as opposed to bilateral amputees. However, Toth...
(1991) and Heckathorne et al. (1995) showed that the findings of Keller et al. (1947) held regardless of whether the dominant or nondominant hand was used.

The Rancho Los Amigos Easy-Feed Hand (Lansberger et al., 1998) is an example of a new mechanical hand design that is based on this observation that prosthetic hands for persons with unilateral amputations tend to be used to hold objects placed into them with the sound hand. The Easy-Feed Hand is a mechanical children’s hand that is easy to push objects into but difficult to pull things out of, making it good for grasping or hanging onto objects. A version of this concept is being commercialized as the Live Touch hand from TRS (Boulder, Colo.).

32.4.3 Hand Width-of-Opening

Most manipulations of the hand are precision manipulations of the palmar prehension kind where the thumb directly opposes the index finger and/or the middle finger. In this mode, most of the hand’s actions are performed with a hand opening of about 5 cm (2 in) (Keller et al., 1947). When designing for dominant hand function, palmar prehension is the desirable pattern limited emphasis on wide opening. For nondominant hand function, where the hand is used essentially as a portable vice with objects being placed into it, a wide opening becomes more important. From a design perspective, allowing the hand mechanism to open 10 cm (3.5 to 4 in), instead of 5 cm (2 in) enables the mechanism to perform the cylindrical prehension power grasp with minimal extra design effort. In general an artificial hand should be able to open at least 10 cm (3.5 to 4 in), or enough to grasp a beverage can or a Mason jar.

32.4.4 Passive Adaptation During Grasping

The grip of the hand is improved by the ability of the hand to passively adapt to the shape of an object grasped. A grasped object depresses, or indents, the skin and underlying soft tissues of the hand, at first meeting little reaction force. Consequently, the soft tissue adapts easily to the shape of the object grasped. However, the mechanical properties of the soft tissue are nonlinear, and the conforming tissue becomes more rigid as pressure is increased. The rise in tissue stiffness after conformation to shape enables objects to be grasped securely. This feature of the human hand would seem to be useful for robotic and prosthetic systems. In prosthetics, the passive adaptability afforded by the soft tissue of the hand is mimicked, to some extent, by lining the prosthesis mechanism with a soft plastic and covering it with a cosmetic glove. In robotics, it is common to use a compliant coating on an end effector to stabilize a robot arm during contact with hard surfaces.

32.4.5 Non-Hand-Like Prehensors

The reduction of most prosthetic terminal devices to a single degree of freedom was a compromise to make the best use of the available control sources. A common transhumeral (above-elbow) body-powered prosthesis has two control cables (two active DOFs), one for an elbow lock to switch the control of the other control cable from the elbow to the terminal device. The terminal device is generally a split hook. A split hook is used when maximum function is desired (Fig. 32.19). Although a split hook is a single DOF device, a split hook can reproduce tip, lateral, cylindrical, or hook prehension depending on which part of the hook is used, making it a very simple and versatile device. This is a contributing factor to the success of body-powered prostheses over externally powered prostheses. Another contributing factor is the “thin” nature of the hook fingers, which only minimally restrict the user’s vision of the object being manipulated.

The use of split hooks highlights the trade-off made between form and function. The hook bears little resemblance to the natural hand but is widely used because of the function it affords if only one DOF is available for terminal device control. Split-hooks are available in many variations from
Hosmer-Dorrance Corp. and Otto Bock Healthcare. Fryer & Michael (1992) provide a thorough review of the various types of hooks currently available.

In an effort to capitalize on the function of the split hook the Hosmer-Dorrance NU-VA Synergetic Prehensor uses a split hook in an externally powered configuration. Other commercially available, externally powered, non-hand-like prehensors include the Otto Bock Greifer and the RSLSteeper Powered Gripper. The NU-VA Synergetic Prehensor, the Otto Bock Greifer, and the Steeper Powered Gripper incorporate many of the previously mentioned prehension patterns (Fig. 32.20).

32.4.6 Hand-Like Prehensors

The de facto standard externally powered hand-like prosthesis is the single-DOF Otto Bock System Electrohand (Fig. 32.12). When used in a prosthetic fitting, a plastic hand-form liner is pulled over the mechanism and a PVC or silicone-rubber cosmetic glove is then pulled over the liner. This gives the hand good overall static cosmesis at the expense of reduced overall mechanism performance. The liner and cosmetic glove act as springs to oppose the opening of the hand by the mechanism, thus degrading the overall performance of the hand. De Visser and Herder (2000) advocated the use of compensatory mechanisms to reduce the effect of the liner and glove on the mechanism’s performance.

RSLSteeper Ltd. (Roehampton, England), and Centri (Sweden) also manufacture single-DOF devices for the adult. Single-DOF child-size hands are also available from Variety Village, Otto Bock Orthopaedic, Inc., and RSLSteeper Ltd., among others. Michael (1986) provides a nice overview of the commercially available powered-hand mechanisms of the day, whereas Heckathorne (1992) provides in-depth descriptions and technical specifications of all the externally-powered components available at the time of writing.

To be clinically viable, the fingers and thumb of most prosthetic hands are nonarticulated and have a single axis of rotation. This minimizes the number of moving parts, reduces complexity and increases robustness. In these handlike prehensors palmar prehension is achieved by single-joint fingers that are fixed in slight flexion at a position approximating the interphalangeal joint. The resulting finger shape also creates a concave inner prehension surface that can be
used to provide cylindrical prehension (Heckathorne, 1992). A single joint thumb opposes these fingers by rotating in the same plane as, but in the opposite direction to, the fingers. All these mechanisms are typically used in a prosthesis that has no wrist flexion or extension. This can be a problem when trying to pick up small objects from a surface. The fixed wrist combined with poor line of sight of the object to be grasped can lead to nonphysiological movements resulting in poor dynamic cosmesis.

**32.4.7 Planes of Motion of the Thumb**

The physiological thumb has two axes of rotation, the metacarpophalangeal (MP) joint and the carpometacarpal (CP) joint giving it many degrees of freedom. Blair and Kramer (1981) claim that the thumb accounts for up to 40 percent of the function of the hand. However, prosthetic hands only have a single axis thumb. Lozac’h (1984; Vinet et al., 1995) performed a series of experiments to find if there was a preferred working plane for a single-DOF active thumb. From these experiments he determined that the preferred working plane of the thumb lay between 45 and 55 degrees (Fig. 32.21). These findings conform to those reported by Taylor (1954) who, from a statistical analysis on natural unrestricted prehension, concluded that: “in palmar prehension, the thumb approaches the fingers in a plane inclined approximately 45 degrees to the palmar plane.”

This finding was implemented in a single-degree-of-freedom, multifunctional hand design (Lozac’h et al., 1992; Vinet et al., 1995). Because the hand had articulated fingers that could move independently of each other it was capable of forming an adaptive grip with which to grasp objects. An adaptive grip meant that it required much lower forces than conventional prosthetic hands to hold objects. Lozac’h et al. (1992) also found that the hand reduced the number of arm and body compensatory movements during both the approach and utilization phases of prehension, as well as greatly improving object visibility, prehension cosmesis (dynamic cosmesis) and grip stability, particularly for large objects. Unfortunately, they also found that they had to further improve the design for greater durability and long-term reliability.

The finding that the preferred working plane of the thumb lay between 45 and 55 degrees is not reflected in many other prosthetic or orthotic devices. Ken worthy (1974) designed a hand in which the thumb moved at 90 degrees to the fingers, i.e., the hand used a side-pinch type of grip (lateral prehension). This hand was developed for use with the CO₂-powered arms (Simpson, 1973) of the Orthopaedic Bio-Engineering Unit (OBEU) of Princess Margaret Rose Orthopaedic Hospital, Edinburgh, Scotland.

The motivation for placing the thumb at this angle came from trying to improve the visibility of the object to be grasped during the final stages of the grasping process and to improve flat-surface operation. It was an attempt to overcome the poor dynamic cosmesis that results from the use of a traditional pincer-grip prostheses (such as the Otto Bock System Electrohand) with a rigid wrist.

Traditional pincer-grip prostheses are easy to use when picking up tall objects (greater than about 25 mm above the surface), but the only way that low objects (less than 25 mm) can be grasped is by approaching the object from above. This requires the amputee to maintain an unsightly line of attack.
(poor dynamic cosmesis) in order to see the object being grasped and to permit flat surface operation. This unusual motion, in turn, draws attention to the user.

Kenworthy’s hand had the fingers fixed with only the thumb being able to move. While flat-surface operation is important, it is perhaps not the most important feature of prosthesis design. This hand was designed primarily for use in bilateral arm systems in which one hand was a conventional pincer-type hand and the other was of Kenworthy’s design. The 90 degrees of the Kenworthy hand was chosen so that at least one hand of the bilateral arm system could have flat-surface operation.

Another hand, developed by Davies et al. (1977) also at the OBEU, was a monofunctional body-powered device in which the thumb moved about an axis inclined at an angle of 60 degrees to the middle and index fingers. This hand, called the OBEU Hand, was a compromise between the more traditional pincer-type of hand and the Kenworthy hand. The motivation for this hand was to retain the pincer-type function while improving the overall visibility of the object to be grasped. Additionally flat-surface operation could be achieved by allowing the fingers and thumb to sweep the surface. In this design both the thumb and fingers moved simultaneously. The 60 degrees for the OBEU hand was chosen because it allowed the loci of the tips of the thumb and index and middle fingers to move approximately in a horizontal plane when the wrist axis was at about 25 degrees to the horizontal so that they could sweep the surface. More recently the hand under development by the WILMER group (Plettenburg, 2002) also has its thumb at an angle of 45 degrees to the fingers. All these designs were an attempt to improve the visibility of the object being grasped and consequently the dynamic cosmesis of the prosthesis while using single-DOF handlike mechanisms.

32.5 MULTIFUNCTIONAL MECHANISMS

There have always been attempts to design fully articulated arms and hands in effort to recreate the full function of the hand. In prossthetics, it was not until after the Second World War and in particular during the 1960s and 1970s that much time, effort, and money was invested in the development of externally powered multifunctional hand-arm systems. Much of the impetus for the research of the 1960s was as a result of the drug thalidomide that was prescribed to a large number of pregnant women in Europe, Canada, and Australia. The drug acted on the fetus in such a way as to inhibit development of the limbs, causing the child to be born with one or more fetal-sized limbs that never developed further.

Worldwide, many new government initiatives were established to help in the development of externally powered, multifunctional, complete arm systems for these children. Prime among these being the Edinburgh Arm (Simpson, 1969), the Boston Arm (Mann, 1968; Mann & Reimers, 1970), the Philadelphia Arm (Taylor & Wirta 1970; Taylor & Finley, 1974), the Wasada Hand (Kato et al., 1970), the Belgrade Hand (Razic, 1973; Stojiljkovic & Saletic, 1975), the Sven Hand (Herberts et al., 1978), the Utah Arm (Jacobsen et al., 1982).

While many ideas were tried and tested during this period, only a few devices ever made it from the laboratory into everyday clinical practice. The Edinburgh Arm, which was pneumatically powered, saw some clinical use, but it was too complex and as a result lacked robustness making it susceptible to breaking down. While this arm is not available today, it is important because it was an implementation of Simpson’s ideas on extended physiological proprioception (EPP). The Boston Arm, developed at MIT, was the first myoelectrically controlled elbow. This elbow was extensively redesigned (Williams, 1989) for commercial production. The current version is the Boston Elbow III (Liberating Technology, Inc.). The Utah Arm is commercially available through Motion Control, Inc. (Sears et al., 1989).

The Sven Hand never found widespread clinical use, even though it was extensively used in multifunction control research using pattern recognition of myoelectric signals (Lawrence and Kasdors, 1974). Henry Lymark, the director of the Handikappinstitutet of Stockholm, Sweden, later created a simplified version of the Sven hand called the ES hand. This hand was an attempt to produce a more robust and hence more clinically viable version of the Sven hand. The ES hand possessed an adaptive grip and a passive two-position thumb. It was powered by a single motor with
differential drives to the fingers. Unfortunately, Lymark died soon after the initial development of this hand and the project was never continued.

The Philadelphia Arm of Taylor and Wirta (1970) and Taylor and Finley (1974) also never found clinical use, but like the Sven Hand found use as a research tool for multifunction control using weighted filters for the pattern recognition problem. The Belgrade hand too was never used clinically but has ended up in the robotics field in the form of the Belgrade/USC robotic hand (Beattie et al., 1994).

In the end, most multifunctional prosthesis designs are doomed by practicality, even before the control interface becomes an issue. Prostheses users are not gentle with their devices; they expect them to work in all sorts of situations never dreamed of by their designers. Most mechanisms fail because of poor durability, lack of performance, and complicated control. No device will be clinically successful if it breaks down frequently. A multifunctional design is by its nature more complex than a single-DOF counterpart. From a maintenance standpoint this means the device will have more components that are likely to fail. Articulated joints on fingers are more likely to fail than monocoque, or solid finger, designs. However, a compromise must be reached if increased function is to be achieved. Some of the robustness and simplicity of a single-DOF device must be traded to achieve the increase in performance possible with a multi-DOF hand.

Another practical consideration is performance. The hand must be able to generate enough torque and speed and have a sufficient width of opening to be useful to the user. Many devices have been designed in the laboratory that have insufficient performance once a cosmetic glove is added. A cosmetic glove is standard for prosthetic hands, and unless a mechanism is specifically designed not to require a glove, the effect of the glove on performance must be taken into consideration. The ES Hand was designed to work in its own cover. The Belgrade Hand needed an additional cover. The pinch force of a multifunctional hand does not have to be as high as that possible with current commercially available single-DOF hands because the adaptive nature of the grip enables objects to be encompassed within the hand. But they should still be capable of high speeds of opening and have a pinch force of at least 68 (15 lbf) in accordance with Peizer et al. (1969).

Most of the early artificial arms for the high-level amputee were designed as complete arm systems. But as these systems failed to provide the expected function, there was a move away from designing complete arm prostheses to the design of specific or modular components, such as externally or body-powered elbow joints, powered or passive wrist rotators, passive humeral rotators, and whole ranges of different hooks, hands, and prehensors. The current trend is to use a “mix and match” approach to optimize the function available. This modular approach has the advantage of providing great flexibility and practicality for system design. However, it will probably never be able to attain the high functional goals that may be possible from a more integrated standpoint. Many of today’s commercially available externally powered elbow systems (Boston Elbow, Liberating Technology, Inc; NYU Elbow, Hosmer-Dorrance; and Utah Elbow, Motion Control) owe their origins to this era of upper-limb research.

### 32.5.1 Is There an Optimum Number of Degrees of Freedom for Multifunctional Hands?

To be clinically viable a multifunctional hand must be robust. This was one of the reasons why the Sven Hand was simplified to the ES Hand—so that it might find some clinical application. A possible compromise to the dilemma of robustness versus increased function possible with a multi-DOF hand is to limit the device to those degrees of freedom necessary to replicate Keller et al. (1947) grasp patterns. This idea of providing sufficient DOFs to recreate Keller et al. grasp patterns turns up in many unrelated fields when compromise must be made between function and some other variable.

Professional scuba diver gloves trade function for warmth in order to extend dive times (Fig. 32.22). A mitten is warmest, whereas a glove with individual fingers is the most functional. Professional scuba diver gloves are a compromise, having the thumb and index fingers free and the middle, ring, and little fingers together. This configuration affords the diver the basic prehension patterns of the hand while at the same time keeping the bulk of the hand warm.
In the area of remote manipulation, the SARCOS system (Jacobsen et al., 1990) uses a three DOF hand for the slave manipulator terminal device and limits the hand of the operator to the same three DOFs when controlling the master arm. Constraining the operator's hand to the same DOFs as the slave, and vice versa, enables the operator to extend his or her proprioception into the remotely controlled terminal device. Forces experienced by the slave are reflected back to the master and experienced by the operator. In this mechanism, the thumb has two DOFs while the three fingers (middle, ring, and little) have the third DOF. The index finger was kept rigid, providing a stable platform against which to operate.

For space suit gloves, the Direct-Link Prehensor (Direct-Link Prehensor, 1991a&b) limits the motions of the operator’s hand to three DOFs. Space suit gloves are bulky and stiff due to the suit’s pressurization. This stiffness results in limited external dexterity and excessive hand fatigue. Also, as in the case with diver’s gloves, tactile sensation and manual dexterity are lost because the hand is gloved. The Direct-Link Prehensor is a two finger device with the thumb mounted at 45 degrees to the fingers similar to the work of Lozac’h (1988).

In the area of surgery, Beasley (1983) described a surgical procedure to provide a functional four-DOF hand for persons with C5–C6 quadriplegia. This procedure makes possible precision prehension with careful positioning of the stabilized thumb to oppose the actively flexed index and middle fingers. The result is a functional hand that retains some of its sense of touch.

In the field of functional electrical stimulation (FES), surgical techniques similar to those of Beasley (1983) have been combined with implantable four-channel FES electrodes to enable patients with flail arms to reproduce the palmar and lateral grasp patterns of Keller et al. (Triolo et al., 1996). In each of these examples either tactile sensation (feedback) was compromised [by gloves or remote nature of the terminal device] or muscular function (control) was impaired. In all cases enabling the hand to recreate Keller et al.'s prehension patterns optimized hand function vs. the number of available control sources. This observation suggests that a prosthetic hand capable of implementing Keller et al.'s patterns would also optimize function versus available control sources (assuming sufficient control sites can be found).

Initially it would appear that such an artificial handlike prehensor would require three or four DOFs to adequately reproduce all six prehension patterns: at least one, more usually two, for the thumb; one for the index finger; and one for the middle, ring and little (MRL) fingers which are combined to move as a single unit. Limiting the thumb to a single degree of freedom and orienting it such that it operates along its preferred plane, 45° (Lozac’h, 1988, Vinet et al., 1995), reduces the number of DOFs to be controlled to three.

Furthermore, if one considers the role of the thumb during grasping it becomes apparent that the prehension pattern adopted by the hand can be made to be a function of the timing (or speed) of thumb closure with respect to the timing (or speed) of finger closure. If a “close” command to the thumb and fingers results in tridigital palmar prehension, in which the thumb engages both the index finger and MRL finger unit, then delaying closure of the thumb a fraction (or reducing thumb speed) will result in tip prehension, in which the thumb only engages the index finger, because the MRL finger unit will pass and miss the thumb (due to the delay) to close on themselves. Delaying thumb
closures more will result in lateral prehension, in which the thumb closes on the side of the index finger, because the thumb will engage neither the index finger nor MRL finger unit and both will close on themselves. Power grasps result from digit shape and a wide width of opening.

A system of this kind, where the timing, or speed, of thumb is controlled, can be configured so a single “open” signal drives all digits (fingers and thumb) back to their start positions. But two “close” signals are needed, one for the index and MRL finger drives and a second for the thumb drive. This implies a one and one-half degree-of-freedom system. If a wiffle tree structure, or a differential drive, is used to drive the fingers, then a single motor could be used to drive all the fingers. It should be noted that prehension (grasping) should not be confused with manipulation. For dexterous manipulation many more degrees of freedom of control are required.

The new Gung-Ho Grip™ for G.I. Joe comprises a 2-DOF hand in which the index finger (trigger finger) is free to move by itself while the middle, ring, & little (MRL) fingers move together. The thumb is rigid and at an angle to the palmar surface of the hand (Lee Publications, 2001).

32.6 SAFETY

Safety considerations are an integral part of any design that has a person in the loop. Limit switches, or a suitable substitute (current sensing), should be used to shut off the motor at the limits of a component’s range of motion (Fig. 32.17). These limit switches should be backed up by mechanical stops capable of physically stopping the component’s drive should the electrical limit switches fail. In this way, while the drive may burn-out, should it run up against the mechanical stops due to a limit switch failure, the user remains safe. Manual overrides for mechanisms should also be included so that if a prehensor fails while it is gripping an object the user can still open it manually. Motion Control has a manual override built into the on-off switch of its new hand. Otto Bock has a break away mechanism that consists of a friction brake. In the Otto Bock mechanism there are multiple interdigitated plates that are preloaded by way of a set screw. The problem with the Otto Bock mechanism is that it will protect the drive train from high external loads by slipping at a preset friction, but it must be screwed tight enough that it cannot be manually unloaded.

32.7 CONTROL

Although the physical design constraints of weight, volume, and power are severe, they are not so severe that multifunctional arms and hands cannot be built that would be of acceptable weight and size. The real problem is, as has been alluded to before, the issue of how to interface a multifunctional arm or hand to an amputee in a meaningful way. It is for this reason that upper-limb prosthetics is often dominated by consideration of control. That is, how can the prosthesis be controlled in such a fashion that it will be an aid rather than a burden to the user?

Childress (1992) presented the following attributes of prosthesis control as desirable. While some of these attributes may be difficult, if not impossible, to achieve in practice, they are still seen as desirable goals.

1. **Low mental loading or subconscious control.** The prosthesis should be able to be used without undue mental involvement. The prosthesis should serve the user; the user should not be the servant of the prosthesis.
2. **User friendly or simple to learn to use.** Any device should be intuitive and natural. An amputee should be able to learn to use the prosthesis quickly and easily.
3. **Independence in multifunctional control.** Control of any function or degree of freedom, should be able to be executed without interfering with the other control functions of a multifunctional prosthesis.
4. **Simultaneous, coordinated control of multiple functions (parallel control).** User should have the ability to coordinate multiple functions simultaneously in effective and meaningful ways without violating the first and third attributes.
5. *Direct access and instantaneous response (speed of response).* All functions, if possible, should be directly accessible to the user, and these functions should respond immediately to input commands.

6. *No sacrifice of human functional ability.* The prosthesis should be used to supplement, not subtract, from available function. The control system should not encumber any natural movement that an amputee can apply to useful proposes.

7. *Natural appearance.* Movements that appear mechanical in nature attract unwanted attention in social situations and may not be pleasing to the eye.

As one might imagine, the level of amputation has very important consequences for the control of a prosthetic device. The level of amputation determines the number of control sites available versus the number of control sites that are needed. This is an inverse relationship. That is, the higher the level of amputation, the fewer are the available control sources but the greater is the amount of function that must be replaced. A control source is the means used by the amputee to control a specific function, or degree of freedom, of the prosthesis, e.g., opening and closing of an artificial hand. The fewer the number of independent control sources, the greater are the number of compromises that must be made in order to achieve a clinically practical device.

Practical inputs typically come from muscular activity, (1) directly, (2) indirectly through joints, and (3) indirectly from by-products of muscular contraction (myoelectricity, myoaoustics, muscle bulge, and mechanical/electrical impedance). Although signals can be obtained from brain waves, voice, feet, eyes, and other places, these sources of control have not been shown to be practical for artificial limb control (Childress 1992).

The primary sources of control for body-powered devices are biomechanical in nature. Movement, or force, from a body joint or multiple joints is used to change position, or develop a force/pressure that can be transduced by a harness and Bowden cable and/or mechanical switches. Typically, inputs such as chin and hand force/movement, glenohumeral flexion/extension or abduction/adduction, bicipital and scapular abduction, shoulder elevation and depression, chest expansion, and elbow or wrist movements are used. However, direct force/motion from muscle(s) has also been used by way of surgical procedures such as muscle tunnel cineplasty (Sauerbruch, 1916) and the Krukenberg cineplasty (Krukenberg, 1917).

For externally powered devices, electromechanical switches and myoelectric control are the main sources of control. The usual input to most externally powered prosthetic components is the plusminus drive wires of the dc motor used to drive the component. As long as the motor receives a voltage of either polarity across these wires, it will run in one direction or the other. This voltage can be an on/off type voltage in which the motor is fully “on” or fully “off” (switch control). Or it can take the form of a variable dc level voltage, in which case motor speed is proportional to the dc voltage level (proportional control). Or it can be a stream of pulses in which the motor drive voltage level is proportional to the amount of “on” time of the pulses (pulse width modulation or PWM). PWM also is a type of proportional control. In all instances, a control source is needed to transduce some body motion or biological artifact into a voltage to drive the motor. While biomechanical inputs are used most extensively in the control of body-powered prosthetic components, the same inputs can be used, through the use of appropriate electromechanical switches or force transducers, to control externally powered components. Fig. 32.23 shows a high level bilateral hybrid fitting that highlights many of these issues.

### 32.7.1 Comments on Proportional Control and Pulse Width Modulation

In proportional control, the amount/intensity of a controlled output variable is directly related (proportional) to the amount of the input signal. For example, the output speed of a dc motor is proportional to the amount of voltage applied to its terminals. This is why dc motors are said to be *speed controlled.* This is also the reason why most of today’s commercially available prosthetic components are speed controlled—it is simple. Output speed is proportional to the amount of input signal. Proportional control is used where a graded response to a graded input is sought.
FIGURE 32.23 This photograph shows a person with bilateral shoulder disarticulations who has been fitted bilaterally using hybrid prostheses. The right side is fitted with a body-powered, single-control cable, four-function system. This is a sequential control system in which the control cable controls elbow flexion-extension, wrist rotation, wrist flexion or terminal device opening and closing depending on which components are locked or unlocked by the chin-actuated mechanical switches. Chin-actuated levers lock and unlock a shoulder flexion-extension unit, a humeral rotator, an elbow, and a wrist rotator. A lever on the wrist locks and unlatches a wrist flexion unit. The terminal device is a voluntary-opening split hook. Shoulder flexion-extension and humeral rotation are under gravity control once they are unlocked. The left side, or electric-powered side, still uses mechanical chin actuated levers to lock and unlock a shoulder flexion-extension joint and a humeral rotator, otherwise the remaining components are externally powered. The arm has an electric-powered elbow and a wrist rotator that are controlled by chin movement against two rocker switches. Pulling on a linear potentiometer actuates the powered prehensor. Output speed, or force, of the prehensor is proportional to how much the transducer is pulled. For this system, the body-powered side is used as the dominant arm, with the electric-powered side assisting in a nondominant role. The prostheses are used for activities of daily living. This fitting highlights a number of important issues that are discussed further in the text.
In position control the position of the prosthetic joint is proportional to the input amount/intensity. The input amount/intensity might be the position of another physiological joint or a force level. If the position of another joint is used as the input then the system is known as a position actuated, position servomechanism. If the amount of force applied by some body part is the input, then the system is a force actuated, position servomechanism. An example is the power steering of a car. Here, the position of the steering wheel is related directly (proportional) to the position of the front wheels. Such a system is an example of a position follower (the position of the wheels follows the position of the steering wheel) or a position servomechanism.

If a further constraint is added whereby the input and output are physically (mechanically) linked such that change in position of the output cannot occur without a change in position of the input and vice versa, then as is the case in the power steering example, the system becomes what Simpson (1974) called an unbeatable position servomechanism or a system that has inherent extended physiological proprioception (EPP). Unbeatable servomechanisms are a subset of position servomechanisms as a whole where the input and output must move together. One cannot beat the other.

With position control the amputee’s ability to perceive and control prosthesis position is directly determined by his or her ability to perceive and control the input signal. A major disadvantage of position control is that, unlike velocity control, it must maintain an input signal to hold an output level other than zero. This means that power must be continuously supplied to the component to maintain a commanded position other than zero. This is one of the reasons why speed or velocity control is the dominant mode of control in externally-powered prosthetics today, despite the fact that it has been shown that position control for positioning of the terminal device in space is superior to velocity control (Doubler & Childress, 1984a&b). Equally, it has been observed that velocity control may be better suited to the control of prehension (Carlson & Primmer, 1978; McKenzie, 1970) but this observation may be due to the poor performance of the prosthetic hand mechanisms of the day and consequently the manner in which amputees tended to use them.

Although proportional control for externally powered prostheses has been around since the 1970s it is only comparatively recently that proportional controllers have become more widely accepted. This is so because of the poor speed and speed of response of earlier systems. If a mechanism is slow, a user will tend to drive it using either a full open signal or full close signal regardless of the type of controller, i.e., slow devices will be used in essentially a switch or “bang bang” mode. It is the bandwidth of the mechanism and controller together that determines speed of response. If the mechanism is fast but the controller introduces delays due to processing, then the speed of response of the overall system will be limited by the controller. However, it is only recently that any prosthetic components have had sufficiently high mechanical bandwidth that users could perceive or have a use for a proportional-type control. The Hosmer-Dorrance NU-VA Synergetic Prehensor, Otto Bock System Electrohands, the Motion Control Hand, The Utah Elbow, and the Boston Elbow III all benefit from the use of proportional controllers. For some users, these devices need a proportional controller so that they can be controlled in a meaningful way because they are too fast to be operated effectively in a switch control mode.

In this day of digital circuits and microprocessor based controllers pulse width modulation is the preferred method of supplying a graded (proportional) control signal to a component. A PWM stream only requires a single digital output line and a counter on the microprocessor to be implemented, whereas a conventional analog signal (linear dc voltage level) requires a full digital-to-analog (D/A) converter (Fig. 32.24). PWM techniques are used extensively in switched-mode power supply design and audio amplifiers (Israelsohn, 2001) and as such, there is a large array of resources available to the designer to choose from.

As implemented on most microprocessors, a period register is loaded with a base period (1/PWM frequency; PWM frequency is usually in the range of 30 kHz for switched mode power supplies but can be as low as 300 Hz in some motor control applications) that is some fraction of the microprocessor’s clock frequency (D’Souza and Mitra, 1997). A duty-cycle register is loaded with a percentage of the base period that corresponds to the desired proportional output voltage level as a percentage of the maximum possible output voltage. At the beginning of each period, the output at the PWM port is high, and the microprocessor counts down until the duty cycle register is zero, at which time the PWM port goes low. The microprocessor continues to count down until the base
A variation on PWM is pulse position modulation (PPM), also known as pulse period modulation or pulse frequency modulation (PFM). In this case, the duty-cycle pulse remains on for a fixed time while the base period is varied. The frequency of the pulses (how close together the pulses are) determines the voltage level. The neuromuscular system is an example of a pulse position modulation system. A muscle is made up of many discrete motor units. A motor unit has an all or nothing response to a nerve impulse in much the same way that a nerve impulse is a nonlinear (thresholded) all-or-nothing event. The level of sustained force output of a motor unit is dictated by the frequency of incidence of the nerve impulses, with the motor units’ dynamics [mechanical properties—inertial and damping properties (acts as a mechanical filter)] holding the force output smooth between incoming impulses. The motor unit is pulse frequency modulated by the nervous system.

Finally, it is possible to have a combination of PWM and PPM where one has variable width pulses and variable periods. In all cases, the output PWM stream is a series of pulses in which the dc voltage level is proportional to the ratio of the amount of on time with respect to the amount of off time for the pulses in the stream. Myopulse modulation, as implemented by Childress (1973) in his myoelectric controller, is an example of this form of pulse modulation (Fig. 32.25). This is an elegant proportional controller that can be implemented with a minimum of components. Because percentage myoelectric signal on time (EMG signal level above an arbitrary threshold) is monotonically related to muscle output force level, a proportional signal can be achieved very simply. This controller
FIGURE 32.25 Myo-pulse modulation (Childress, 1973)—the voltage level seen at the motor is proportional to the ratio of “ON” time of the pulses to “OFF” time. The mean of this ratio is in turn proportional to the intensity of muscle contraction and as such is a very simple means of providing proportional myoelectric control. (a) The trace $\gamma(t)$ shows the switching response of a comparator in response to a time-varying EMG signal $e(t)$ and is given by the expression:

$$\gamma(t) = \frac{V}{2} \left[ 1 + \text{sgn}(|e(t)| - \delta) \right]$$

where $|e(t)|$ is the absolute magnitude of the amplified myoelectric signal $d$ is the threshold, and $V$ is the power supply voltage. The thresholds for switching are $+\delta$ and $-\delta$. One method of choosing these thresholds is to set them at levels corresponding to $\pm 3$ standard deviations of the quiescent (resting) signal amplitude. (b) The schematic shows the key components needed to implement this kind of control. The advantages of this EMG processing approach are (1) the simple electronic implementation, (2) No electrical time constant delay (processing delay)—implies faster response, (3) Control signal already in pulse modulation form, and (4) Wide dynamic range of muscle output.
simply converts a differentially amplified myoelectric signal directly into a pulse stream through the use of a pair of comparators. This pulse stream is then fed directly to the motor H-bridge without further processing. This controller is able to present a pulse stream to the drive motor in advance of the operator being able to detect motion by the innervated muscle, making it ideally suited for prosthetic mechanisms that require a quick speed of response. This controller is now commercially available through Hosmer-Dorrance Corp.

To recover the analog voltage level from a pulse stream requires the use of a low-pass filter (Palacherla, 1997). When used to drive a motor (via an H-bridge), the inertia and damping of the armature in the motor form a low-pass mechanical filter that provides the filtering necessary to smooth the pulses into a continuous signal. A simple RC circuit can be used to smooth the pulse stream if it is not being fed directly into a dc motor. If using a simple RC circuit, the cutoff frequency should be about two decades below the PWM frequency if a smooth output signal is sought. An RC filter is a first-order filter that has an attenuation of 20 dB/decade. Higher-order filters could, of course, be used with cutoff frequencies closer to the PWM frequency. For motors, a smooth signal is not vital.

32.7.2 Myoelectric Control

Myoelectric control derives its name from the electromyogram (EMG), which it uses as a control input. When a muscle contracts, an electric potential (the EMG) is produced as a by-product of that contraction. If surface electrodes are placed on the skin near a muscle, they can detect this signal (Fig. 32.26). The signal can then be electronically amplified, processed, and used to control a prosthesis. While the intensity of the EMG increases as muscle tension increases, the relationship is a complex nonlinear process that depends on many variables, including the position and configuration of the electrodes (Heckathorne and Childress, 1981). Although the EMG is nonlinear it is broadly monotonic, and the human operator perceives this response as more or less linear.

The first externally powered prosthesis was a pneumatic hand patented in Germany in 1915. Drawings of this hand and possibly the first electric hand were published in 1919 in *Ersatzglieder und Arbeitshilfen* (Borchardt et al., 1919). The first myoelectric prosthesis was developed during the early 1940s by Reinhold Reiter. Reiter published his work in 1948 (Reiter, 1948), but it was not widely known, and myoelectric control had to wait to be rediscovered during the 1950s. Reiter’s prosthesis

![FIGURE 32.26 Muscle as a biological neural amplifier. The muscle in effect acts as a biological amplifier for the neural signal. The myoelectric signal caused by the neural signal activation of the muscle can then be detected at the skin surface and further amplified electronically for use in logic circuitry.](image-url)
consisted of a modified Hüfner hand that contained an electromagnet controlled by a vacuum tube amplifier. The prosthesis was not portable but was instead intended for use at a workstation, although Reiter did hope that one day it might be portable.

The Russian hand was the first semipractical myoelectric hand to be used clinically. This hand also had the distinction of being the first to use transistors (germanium) to process the myoelectric control signal (Childress, 1985). In this country, following World War II, the Committee on Artificial Limbs contracted with IBM to develop several electrically powered limbs. These were impressive engineering feats in their day but never found use outside the laboratory (Klopsteg and Wilson, 1956).

Myoelectric control has received considerable attention since it first appeared during the 1940s, and there is an extensive body of literature describing the EMG’s characteristics and properties (see Parker and Scott, 1985; Basmajian and DeLuca, 1985). It was considered to be the cutting edge of technology of the day and was advanced as a natural approach for the control of prostheses since it made it possible for amputees to use the same mental processes to control prosthesis function as had previously been used in controlling their physiological limb (Hogan, 1976; Mann, 1968).

Usually the EMG is amplified and processed (bandlimited, rectified, and thresholded) to provide a dc signal that is related to the force of muscular contraction; this is then used to control the prosthesis (Scott, 1984) [Fig. 32.27]. EMG processing in a typical prosthetic myoelectric control system involves two pairs of differential dry metal electrodes and a reference electrode (Fig. 32.28). Although the electrodes are referred to as dry, the environment inside a prosthetic socket causes the amputee’s residual limb to sweat, which creates a conductive interface between the skin and the electrodes (Fig. 32.29). Traditionally, myoelectric control uses electrodes placed on the skin near each of a protagonist-antagonist pair of muscles to control a single degree of freedom. For below-elbow fittings, this usually means electrodes on those muscle groups responsible for flexion and extension of the wrist and fingers (Fig. 32.30). Thinking about flexing or extending the “phantom” fingers controls closing or opening, respectively, of a terminal device.

Surface EMG signal amplitude (RMS) is approximately 100 µV for a moderately contracted forearm muscle. This signal must be amplified to a signal with an amplitude in the range 1 to 10 V before it can used. This implies that a gain of upward of 10,000 is needed. The bandwidth for the surface EMG signals is 10 to 300 Hz, with most of the signals’ energy in and around 100 Hz (Childress, 1992). Differential amplifiers are used to amplify the EMG signal because the small EMG signal is often superimposed on large common-mode signals that, at these gain levels, would saturate an amplifier in a single-mode configuration. A differential amplifier can remove the large
FIGURE 32.28 Schematic of the typical myoelectric processing scheme used in standard commercially available two-site myoelectrically controlled systems. (See also Figs. 32.29 and 32.30). The major difference here from Fig. 32.27 is that there must be some way to decide which of the two incoming signals to use to drive the motor. Usually a drive signal proportional to the magnitude of the difference between the two incoming signals is sent to the motor.

FIGURE 32.29 A standard transradial myoelectric prosthetic interface (socket). The battery pack, wrist unit, and myoelectrodes are all fitted into a custom-made laminated prosthetic socket and forearm. The socket is fabricated after a mold made from the amputee’s residual limb.
common-mode signals, leaving only the potential difference (EMG signal) between the electrodes to be amplified. This has the effect of most effectively amplifying the EMG signal frequencies around 100 Hz. Because of large gain requirements these differential amplifiers are seldom single op-amps but instead use multiple stages to meet the gain requirements.

Once the EMG signal has been amplified and bandlimited, it is then changed into a dc signal by rectification, by squaring or some other appropriate nonlinear processing. This dc potential is then commonly smoothed with a low-pass filter to remove the pulses and extract the envelope of the signal. For on/off, or switch, control, the smoothed dc voltage is then compared in a logic circuit with a threshold voltage. If the signal is greater than the threshold voltage, then power is supplied to the prosthesis motor, otherwise the power remains off. For proportional control, the smoothed voltage is fed to the motor drive circuit.

When used for proportional control, the EMG signal is usually treated as an amplitude modulated signal, where the mean amplitude of the cutaneous signal is the desired output from the myoelectric processor. However, in order to have accurate estimates of muscle force from EMG signals, a processing system with high signal-to-noise ratio (SNR) as well as short rise time (fast response) is required (Meek et al., 1990). Unfortunately, there is a filtering paradox whereby it is possible to have either fast response or high SNR, but not both (Jacobsen et al., 1984). To overcome this perceived problem, Meek et al. (1990) proposed using an adaptive filter in which the time constant of the low-pass filter used in the final stage of EMG processing (acquire signal envelope) was varied depending on the rate of change of the EMG signal. Their assumption was that an amputee, when moving quickly, will tolerate noise (low SNR) but will not tolerate delays in control. When holding the prosthesis steady or performing slow, dextrous tasks such as threading a needle, the amputee will tolerate slow response as long as there is low noise (high SNR). For the time being, this issue is largely academic. The available components do not have sufficient bandwidth for this issue to matter; however the designer needs to be aware of this filtering paradox should they be involved in the design of high-bandwidth myoelectrically controlled systems.

Processing steps take time! Any delay in the response of the output to a change in the input of greater than about 100 $\mu$s is perceptible to the human operator as a sluggish response. Any delay in response reduces the overall system bandwidth. Parker and Scott (1985) suggest that delays between input and output should not exceed 200 $\mu$s, and even this can be unacceptable for use with high performance prosthetic components.

Myopulse modulation (Childress, 1973) offers a means of processing myoelectric signals that maximizes the speed of response of an externally powered component. This technique eliminates the delays that are introduced by analog or digital filtering methods, provides proportional control, and
keeps the component count to a minimum, thus keeping size and power consumption down. Another alternative method of myoelectric signal smoothing, called *autogenic backlash* (Bottomley, 1965), produced a more or less consistent direct current (DC) output from a fluctuating myoelectric signal while not sacrificing time response.

### 32.7.3 Comments on Myoelectrodes

A differential amplifier subtracts out those elements of the measured signal that are common to the two inputs of the amplifier and leaves the difference (Fig. 32.31). For prosthetic myoelectric control applications, each input to the amplifier is some form of button, or bar, electrode (Fig. 32.32) that sits on the skin’s surface. This pair of electrodes should be oriented along the long axis of the muscle from which the EMG is being measured. This is so because the EMG travels along the muscle’s length, creating a difference in potential between the electrodes that forms the input to the differential amplifier. The greater the distance between the electrodes, the larger the potential difference. It should be understood that muscles come in many forms: pennate, bipennate, smooth, etc., but that in the upper limb, the long axis of the superficial muscles, from which the EMGs are generally taken, runs broadly parallel to the underlying skeleton. The situation is complicated at the shoulder disarticulation level, where the direction of the muscles, e.g., the pectoralis or the infraspinatus, are not so obviously aligned to any long bones. In this case electrodes are aligned to the direction of action of the muscle fibers being measured.

The unwanted large common-mode noise signals, on the other hand, are equally present at both electrodes regardless of the orientation of the electrodes. However, these common-mode noise signals are more likely to be the same at both electrodes if the electrodes are located physically close together. However, the potential difference of the actual EMG between the electrodes will be reduced with small spacing. Thus one trades better common-mode rejection (CMRR) for reduced gain. Commercially available myoelectrodes from Otto Bock consist of a pair of bar electrodes spaced about 20 mm apart with a circular reference electrode located between them. The location of Otto Bock’s reference electrode presents problems (Fig. 32.32c). A better configuration, from a CMRR standpoint is the Hosmer-Dorrance electrode pair which consists of button electrodes 11 mm (7/16 in) in diameter spaced about 28 mm (1 1/8 in) apart with a separate reference electrode that is located as far away from any electrodes as possible within the confines of the prosthesis (Fig. 32.32). The output from Otto Bock electrodes is not the raw EMG but rather an amplified, bandlimited, rectified and smoothed version of the EMG, i.e., Otto Bock electrodes output a processed version of the EMG signal. Hosmer-Dorrance electrodes output an amplified and bandlimited signal.

A physically separate and remote reference electrode is desirable because the reference electrode is not electrically isolated from the inputs to the differential amplifier. The reference electrode is in contact with the person and consequently is electrically coupled to both inputs of the amplifier by the electrical impedance of the body. This impedance is a function of distance, i.e., the further away the reference electrode is, the greater is the attenuation. The reference electrode forms a floating ground signal rather than a true ground. Furthermore, the noise signal that can be introduced by the reference electrode is not necessarily a common-mode signal that will be removed by the differential amplification process.

The size of these electrodes could be greatly reduced given the availability today of many low-offset, ultrafast single-supply op-amps for the cellular phone industry. The ultralow offset voltage of these op-amps allows for higher gains to be used. This in turn would facilitate the design of smaller electrode pairs with higher common-mode rejection ratios. Physically smaller electrodes would allow more electrodes to be placed on the same residual limb.

A final comment about safety. The input lines from the electrodes in contact with the skin should be capacitively coupled to the inputs of the differential amplifier. The surface electrodes should not be directly attached to the differential amplifier inputs. The capacitive coupling blocks the flow of dc current from the amplifier to the user in the event of an electronic failure. If capacitive elements are not present to protect the user, then the user can suffer burns of the skin under the electrode site similar to those reported by Selvarajah and Datta (2001) for a myoelectric hand using a RSLSteeper single-site electrode.
FIGURE 32.31 Schematic of how differential electrodes function. Dry button electrodes I and II are placed over a muscle belly on the skin surface in the line of the long axis of the muscle. These electrodes form the positive (I) and negative (II) inputs of a differential amplifier electrode pair. (a) In skeletal muscle, the individual muscle fibers are generally oriented along the long axis of the muscle. A single axon from the central nervous system innervates a single motor unit usually (for skeletal muscle) at the end of the muscle closest to the spinal chord (i.e., the proximal end). The points where the nerve fibers attach to the individual muscle fibers of a motor unit are called “motor end plates.” When a nerve sends a signal to an individual muscle fiber to contract, the electric signal travels in both directions away from the motor end plate. For this reason the electrodes of the differential amplifier should be oriented along the long axis of the muscle’s fibers. (b) Depolarization wave that travels in both directions away from the motor end plate along individual muscle fibers when activated by a motor nerve fiber. (c) Typical signal seen at the output of the differential amplifier for a single muscle fiber. At (i) the depolarization wave is directly under the positive input (I) with nothing under the negative input (II) to the differential amplifier and consequently registers as a positive going wave at the differential amplifier output; at (ii) the depolarization wave registers equally at the positive (I) and negative (II) inputs and is therefore canceled out, giving a zero at the output of the differential amplifier; at (iii) the depolarization wave is now directly under the negative input (II), with nothing under the positive input (I), so this registers at the differential amplifier output as a negative going wave. In reality, there are thousands of fibers all at different stages in their firing process, which results in the quasi-random signal seen at the skin surface.
32.48 REHABILITATION ENGINEERING

Myoacoustic signals are auditory sounds created as a by-product of muscle contraction. A myoacoustic control system is very similar in structure to a myoelectric control system, but the elimination of unwanted acoustic noise appears to be a bigger problem than the elimination of unwanted electrical noise in myoelectric systems (Barry and Cole, 1990). It has not gained widespread use.

Tendon or residual muscle movement has been used to actuate pneumatic sensors when interposed between a prosthetic socket and the superficial tendons and/or muscle. These sensors can be used for prosthesis control. The Vaduz hand, which was developed by a German team headed by Dr. Edmund Wilms in Vaduz, Liechtenstein, following World War II, used muscle bulge to increase pneumatic pressure to operate a switch-controlled, voluntary-closing, position-servo hand. This hand can be considered to be a forerunner of the pneumatic Otto Bock Hands of the 1970s and the electrically powered Otto Bock Hands of today.

Simpson (1966) also used muscle bulge to provide a control signal for the proportional control of an Otto Bock gas-operated hand. The width-of-opening of the hand was proportional to the force applied by the muscle’s bulge. This device was an example of a force actuated position servomechanism, and because the bulging muscle had to achieve a significant pressure to ensure sufficient force for control-valve operation, a subconscious feedback path existed through the skin’s own pressure sensors. This device was a precursor of Simpson’s concept of extended physiological proprioception (EPP) (Simpson, 1974).

In 1999 the Rutgers multifunctional hand (Abboudi et al., 1999) received considerable media attention because the developers reported a multifunctional controller that allowed amputees to play

32.7.4 Myoacoustic Control

Myoacoustic signals are auditory sounds created as a by-product of muscle contraction. A myoacoustic control system is very similar in structure to a myoelectric control system, but the elimination of unwanted acoustic noise appears to be a bigger problem than the elimination of unwanted electrical noise in myoelectric systems (Barry and Cole, 1990). It has not gained widespread use.

32.7.5 Prosthesis Control Using Muscle Bulge or Tendon Movement

Tendon or residual muscle movement has been used to actuate pneumatic sensors when interposed between a prosthetic socket and the superficial tendons and/or muscle. These sensors can be used for prosthesis control. The Vaduz hand, which was developed by a German team headed by Dr. Edmund Wilms in Vaduz, Liechtenstein, following World War II, used muscle bulge to increase pneumatic pressure to operate a switch-controlled, voluntary-closing, position-servo hand. This hand can be considered to be a forerunner of the pneumatic Otto Bock Hands of the 1970s and the electrically powered Otto Bock Hands of today.

Simpson (1966) also used muscle bulge to provide a control signal for the proportional control of an Otto Bock gas-operated hand. The width-of-opening of the hand was proportional to the force applied by the muscle’s bulge. This device was an example of a force actuated position servomechanism, and because the bulging muscle had to achieve a significant pressure to ensure sufficient force for control-valve operation, a subconscious feedback path existed through the skin’s own pressure sensors. This device was a precursor of Simpson’s concept of extended physiological proprioception (EPP) (Simpson, 1974).

In 1999 the Rutgers multifunctional hand (Abboudi et al., 1999) received considerable media attention because the developers reported a multifunctional controller that allowed amputees to play
the piano in a laboratory setting. This controller used multiple pneumatic sensors that were actuated by the movement of the superficial extrinsic tendons associated with individual finger flexors. For this hand to be clinically viable the developers need to resolve some of the issues that led to the failure of the previous attempts at using pressure/muscle hardness transducers. These issues include the system not being able to differentiate between actual control signals from a tendon and external pressures or impacts from objects in the environment. A prosthesis wearer interacting with the real world will exert forces and moments on the socket, which may actuate the pressure sensors and issue commands to the drive system of the prosthesis.

Otto Bock developed a control system using this idea to control an electric prosthesis in the 1970s (Nader, 1970). Otto Bock’s selling points for this system was that it was immune to electrical and magnetic fields, unaffected by changes in skin impedance, and free from a quiescent current drain. This system was used to provide on-off and proportional control. Although the system has been commercially available for a number of years, it’s use is not widespread. Members of the Technical University of Delft investigated a similar system to control a pneumatically operated prosthesis (Gerbranda, 1974; Van Dijk, 1976). They found that although the output signal of a muscle hardness transducer is, in theory, proportional to muscle contraction, that external disturbing forces exerted on the prosthesis were too large to practically employ proportional control. Hence this system could be used only for on-off control.

32.7.6 Neuroelectric Control

Neuroelectric control, or the control of a prosthetic device by way of nerve impulses, would appear to be a natural control approach. Although there is, and has been, research concerning prosthesis connections with nerves and neurons (Edell, 1986; Kovacs et al., 1989; Andrews et al., 2001) the practicality of human-machine interconnections of this kind is still problematic. Nerve tissue is sensitive to mechanical stresses, and this form of control also requires the use of implanted systems.

Edell (1986) attempted to use nerve cuffs to generate motor control signals in experimental systems. Kovacs et al. (1989) explored the use of integrated circuit electrode arrays into which the nerve fiber was encouraged to grow. Andrews et al. (2001) reported on their progress in developing a multipoint microelectrode peripheral nerve implant. They used a 100 x 100 grid array of silicon microelectrodes that they inserted into a peripheral nerve bundle using a pulse of air. This group is still working with animal models but claims to be able to identify individual neuron action potentials and also claims good long-term results so far. Concerns about how permanently electrode arrays are fixed have yet to be addressed.

A variation on the use of peripheral neural interfaces is the use of electroencephalogram signals (EEG’s). These are electric signals that are detected on the surface of the skull and that result as a by-product of the natural functioning of the brain. Reger et al. (2000) demonstrated a hybrid neurorobotic system based on two-way communication between the brain of a lamprey and a small mobile robot. In this case the lamprey brain was kept alive in vitro and was used to send and receive motor control and sensory signals. The lamprey brain was an in vitro preparation that was used to send and receive motor control and sensory signals. While this is a long way from any practical use in prosthetics it demonstrates a bidirectional interface between nervous tissue and a machine. IEEE Spectrum (2001) reported on research at Duke University, where researchers had implanted an electrode array into the cerebellum of a monkey and through the use of appropriate pattern recognition software had it control a remote manipulator at MIT over 1000 km away via the Internet. One of the main practical goals cited for this research is to put paralyzed people in control of their environments. Finally, Kennedy et al. (2000) reported on the use of implanted electrode arrays to provide persons with “shut-in” syndrome control of a computer cursor.

In the area of functional electrical stimulation (FES), Loeb et al. (1998) have designed and built what they consider to be a new class of implantable electronic interfaces with nerve and muscle. These devices, which they call BIONs, are hermetically encapsulated, leadless electrical devices that are small enough to be injected percutaneously into muscles (2 mm diameter by 15 mm long). This group’s second-generation BIONs will be able to send outgoing telemetry of ongoing movements.
Such devices ought to be able to be used to detect EMG signals and could be used for prosthetic control. Also in the area of FES, FDA approval has been granted to implant FES electrodes for the purpose of providing four-DOF control of an otherwise flail limb (Triolo, 1996).

### 32.7.7 Multifunctional Myoelectric Control

For prosthetic arms to be more than just position controllers for portable vices, multifunctional mechanisms that have the ability to have multiple degrees of freedom controlled simultaneously (in parallel) in a subconscious manner need to be developed. Current commercially available multifunctional controllers are generally sequential in nature and take the form of two site, three state multifunctional controllers. Motion Control, Inc., in the ProControl hand-wrist controller, uses rapid cocontraction of the forearm extensors and flexors to switch control between hand opening and closing to wrist rotation. Otto Bock uses a similar control strategy in its wrist-hand controller. Motion Control, Inc., in its elbow controller, uses dwell time (parking) to switch from elbow flexion and extension to hand opening and closure and cocontraction of biceps and triceps to switch control from the hand back to elbow.

An alternative to thinking in terms of controlling individual joints is to think in terms of actual grasp patterns to reduce the number of control sites required. To implement grasp pattern control by way of myoelectric signals, a specific grasp pattern of the hand is tied to a specific pattern of EMG signals that the controller identifies. The premise behind this type of control is that an amputee uses phantom limb sensation to visualize a specific hand grasp pattern, such as palmar prehension or lateral prehension, and consequently fires those muscles, or residual muscles, associated with creating that grasp pattern. The controller detects the resulting EMG signals generated by the residual musculature and through the appropriate use of pattern recognition and signal processing techniques extracts key features from these signals. These key features, in turn, enable the controller to recognize the grasp pattern associated with that EMG pattern, enabling the controller to actuate the appropriate motors to generate the desired hand grasp pattern.

In the Philadelphia Arm (Taylor and Finley, 1974) and Sven Hand (Lawerence and Kadefors, 1974) multiple myoelectric signals were used as control inputs, which were processed using adaptive weighted filters. The weights on the filters were adjusted to tailor the filters to a specific individual. In the Philadelphia arm, the choice of location for the myoelectrodes was based on muscle synergies associated with arm movements instead of the phantom limb sensations used in the Sven hand. One of the findings with the Sven hand was that control using myoelectric signals from an intact limb was superior to control using a residual limb and phantom limb sensation. The developers theorized that this was due to the presence of an intact proprioception system.

At present there is ongoing research into multifunctional myoelectric control of artificial hand replacements using time, frequency, and time-frequency identification techniques to identify features in the EMG signals that describe a particular grasp pattern. Hudgins et al. (1993) and Farry et al. (1993) used neural networks to perform the pattern recognition and feature extraction. Farry et al. (1997) used genetic algorithms. Currently, it is the features of Hudgins et al. (1993) that find the most widespread use. But even so a practical multifunctional hand control algorithm has yet to be developed. The processing time required for these complex signal processing algorithms is nontrivial.

### 32.7.8 Physiologically Appropriate Feedback

Physiologically correct feedback, beyond that provided by vision, is essential if low mental loading or coordinated subconscious control of multifunctional prostheses is to be achieved (Fig. 32.33). When prosthetic arm technology moved to externally powered systems, the control modalities shifted, with the exception of Simpson (1974) and a few others, from the position-based cable control of the body-powered systems to open-loop velocity control techniques (such as myoelectric and switch control). That is, prosthetic technology shifted away from cable inputs, which provide sensory and
proprioceptive feedback, to techniques that provide little feedback to the operator beyond visual feedback.

Simplicity was probably the primary reason why prosthetics shifted to open-loop velocity control. The actuator of choice, the electric dc motor is an inherently rate-controlled device (i.e., its output speed is directly proportional to the input voltage), and it can be readily controlled with on/off switches. This resulted in the primary use of open-loop, velocity-controlled, externally powered prostheses. In addition, a velocity controlled system does not draw power to maintain a particular position. Unfortunately, the user must integrate velocity in order to control position when using velocity control. Constant visual monitoring, due to the essentially open-loop nature of myoelectric control, is therefore required for effective operation. For the control of multiple degrees of freedom this places excessive mental load on the user, greatly diminishing any benefits a prosthesis of this complexity might offer. Visual and auditory feedback are slower, less automated, and less programmed than normal proprioceptive feedback and therefore place a greater mental burden on the user. Such systems use onboard intelligence to automatically respond to some external sensor input. To be effective a user must have confidence in the artificial reflex or else they will not relinquish control to the reflex loop.

FIGURE 32.33 Feedback in human prosthetic systems. Most powered upper-limb prostheses are currently controlled primarily through visual feedback with some assistance from what has been called incidental feedback—whine, prosthetic vibration, socket forces, etc., being examples of incidental feedback; i.e., the motor feedback is incidental rather than by design. Attempts have been made to provide supplementary sensory feedback (SSF) through the use of vibrations of the skin, electrical stimulation of the skin, auditory and visual signals, and other means. However, because these methods are supplementary to the normal sensory feedback paths of the body, they fail to present the feedback information in a physiologically useful manner and consequently have not seen much success. Control interface feedback means that the operator receives information concerning the state of the prosthesis through the same channel through which the prosthesis is controlled. Information concerning prosthetic joint position, velocity, and the forces acting on it is available to the operator through the proprioceptors of the controlling joint. Because feedback through the control interface is usually in forms that are easily interpreted by the user, it can be interpreted at a subconscious level, reducing the mental burden placed on the user. Artificial reflexes are closed loops within the controller/prosthesis mechanism itself that seek to remove the operator from the control loop, and as a result to also remove the mental burden placed on the user. Such systems use onboard intelligence to automatically respond to some external sensor input. To be effective a user must have confidence in the artificial reflex or else they will not relinquish control to the reflex loop.

proprioceptive feedback, to techniques that provide little feedback to the operator beyond visual feedback.

Simplicity was probably the primary reason why prosthetics shifted to open-loop velocity control. The actuator of choice, the electric dc motor is an inherently rate-controlled device (i.e., its output speed is directly proportional to the input voltage), and it can be readily controlled with on/off switches. This resulted in the primary use of open-loop, velocity-controlled, externally powered prostheses. In addition, a velocity controlled system does not draw power to maintain a particular position.

Unfortunately, the user must integrate velocity in order to control position when using velocity control. Constant visual monitoring, due to the essentially open-loop nature of myoelectric control, is therefore required for effective operation. For the control of multiple degrees of freedom this places excessive mental load on the user, greatly diminishing any benefits a prosthesis of this complexity might offer. Visual and auditory feedback are slower, less automated, and less programmed than normal proprioceptive feedback and therefore place a greater mental burden on the operator (Soede, 1982).

In contrast, Simpson (1974) advocated augmented cable control. He coined the phrase extended physiological proprioception (EPP) to indicate that the body’s own natural physiological sensors are used to relate the state of the prosthetic arm to the operator. EPP can be thought of as the extension of one’s proprioceptive feedback into an intimately linked inanimate object. Consider a tennis player hitting a ball with a tennis racquet. The player does not need to visually monitor the head of the racquet to know where it will strike the ball. Through experience, the tennis player knows how heavy and how long the tennis racquet is. He or she knows where in space the head of the racquet is located based on proprioceptive cues
from his or her wrist and hand; that is, how it feels in his or her hand. The tennis player has effectively extended his or her hand, wrist, and arm’s proprioception into the tennis racquet.

The use of locking mechanisms to switch control from one component to the next in a sequential control cable system provides a similar type of control. The prosthesis, when locked, becomes a rigid extension of the user. The user can use this rigid extension to interact with the real world by pulling and pushing on objects. Because the prosthesis is rigid, the user “feels” the forces exerted on the prosthesis and has a sense of feedback about what the prosthesis is doing. The user has extended his or her proprioception into the prosthesis, in a fashion similar to how the tennis racket becomes an extension of the player. When the device is unlocked, it can be easily positioned with minimal effort (Fig. 32.23). The locked/unlocked state of the prosthesis can be thought of as a form of impedance control—two state impedance control. The prosthesis is either in a state of high impedance—rigid or locked, or in a state of low impedance—free or unlocked.

This same principle can be used to provide proprioceptive feedback to a powered prosthetic joint. Consider parking a car that has power steering. The driver feels, through the steering wheel, the interaction between the front wheels and the parking surface, curb, etc. However, the driver does not provide the power to turn the wheels; this comes from the engine. The driver is linked to the front wheels through the steering wheel and has extended his or her proprioception to the front wheels. Essentially, EPP control for externally powered prosthetic components can be thought of as power steering for prosthetic joints.

An ideal EPP system is an unbeatable position servomechanism. It is the mechanical linkage between input and output that converts a simple position servomechanism or velocity controller into an unbeatable position servomechanism. The mechanical linkage closes the loop and provides the path for the force feedback. The mechanical linkage constrains both the input and output to follow each other. In theory, the input cannot get ahead of (or beat) the output and vice versa. In practice, a small error must exist for the controller to operate. This error can be made very small with an appropriate controller gain. Doubler and Childress, (1982a, b), quantifiably demonstrated the potential effectiveness of applying EPP to the control of upper limb prostheses.

However, after being formalized and implemented by Simpson, control of powered limbs through EPP systems has been slow to gain acceptance. This lack of use is possibly due to the harnessing that is required for these systems. If harnessing is required, why not just use a body-powered system. Until a high bandwidth externally powered multifunctional prosthesis that needs parallel control of a number of degrees of freedom is developed, EPP control is unlikely to be rediscovered. Such a prosthesis might be a Boston Elbow III with a powered humeral rotator. In this case, shoulder elevation and depression would control elbow flexion, whereas shoulder protraction and retraction would control humeral rotation.

### 32.7.9 Artificial Reflexes

An alternative approach to the problem of the lack of physiologically appropriate feedback associated with today’s prosthetic components is to automate more of the control functions of a given system. These artificial reflexes strive to reduce both the mental burden placed on the user and the number of control sources required. Artificial reflexes seek to remove the operator from the control loop and use onboard intelligence to automatically respond to some external sensor input and as such, they can be thought of as being similar to an artificial reflex loop. By putting more intelligence into the device through the use of embedded microprocessors, etc., more and more of the decision making process can be automated—requiring less attention on the part of the operator. Artificial reflexes are essentially closed loops within the mechanism/prosthesis itself. This trend of putting more onboard intelligence into prosthetic components is likely to increase in importance in the future. A key to the success or failure of an artificial reflex is that a user must have confidence in the artificial reflex or else they will not relinquish control to the reflex loop.

Otto Bock has a hand that has a microprocessor-controlled slip detector (Otto Bock SensorHand). The thumb has a sensor that detects slippage of an object grasped by the prehensor and automatically
increases prehension force until it detects that the slippage has stopped. Anecdotal evidence suggests
that users of the Otto Bock SensorHand feel more confident that an object they grasp using only
visual cues will be grasped properly because the slip detector prevents the object from falling. The
Otto Bock transfemoral (above-knee) “C-Leg” prosthesis also serves to reduce the mental burden
required of users during walking. This prosthesis uses strain gauges and position transducers coupled
with an embedded microcontroller to ensure that the knee is locked at every step before weight is
borne on the prosthetic leg. Confidence on the part of the user that knee will not collapse if they stop
concentrating on whether the knee is locked at each step will be critical to the success of this device.

Kyberd and Chappell (1994) use a system they call hierarchical artificial reflexes to automate the
control process. In their multifunctional hand, they take the operator out of the loop and use onboard
processing and sensors in the hand to tell the hand what grasp pattern to adopt. The operator only
provides a conventional single degree of freedom open or close EMG signal. The idea is that by
allowing the processor to take control, it reduces the mental loading on the operator. A major factor
in the success or failure of these devices is confidence in the mechanism on the part of the user, so
as to relinquish control to the artificial reflex.

32.7.10 Surgical Innovations Required

Surgeons need to perform innovative procedures that create new kinds of human-prosthesis interfaces
if major advancements in upper-limb prosthesis control are to occur. They will have to create
interfaces between amputee and prosthesis that allow for sensory feedback as well as control signals. Direct muscle attachment is one such interface. The concept of direct muscle attachment is not new.
It had its origins in Italy at the beginning of the twentieth century and was brought into clinical
practice in Germany by Sauerbruch around 1915 (Sauerbruch, 1916). Sauerbruch’s technique, called
muscle tunnel cineplasaty, fashions a skin-lined tunnel through the muscle (released at its insertion)
and enables the muscle’s power to be brought outside the body. Weir (1995; Weir et al., 2001) has
shown the efficacy of tunnel cineplasty control when compared with the control of a conventional
above-elbow, body-powered prosthesis. A similar but alternative surgical procedure called tendon
exteriorization cineplasty (Beasley, 1966) uses tendon transfers combined with skin flaps to bring a
tendon loop outside the body.

The subconscious control possible with an EPP controller operating in conjunction with the
proprioception of a surgically created muscle tunnel cineplasty presents the intriguing possibility of
making independent multifunctional control of a prosthetic hand or arm a reality. Suitable control
muscles and EPP controllers in conjunction with powered fingers of an artificial hand would be a step
toward achieving the goal of meaningful, independent multifinger control of hand prostheses. In a
prototype fitting (Fig. 32.34), a below-elbow amputee with tendon exteriorization cineplasties was
fitted with an externally powered hand that utilized the subject’s exteriorized tendons as control inputs
to an EPP controller (Weir et al., 2001). Movement of the flexor tendon caused the hand to close.
Movement of the extensor tendon caused the hand to open. This was the first clinical fitting of a
powered-hand prosthesis controlled directly by antagonist muscles (via exteriorized tendons) in a
somewhat physiological manner.

The use of muscle/tendon cineplasty procedures would require changes in the way amputation
surgery is performed at the time of initial amputation. Preservation of muscle tone and length is of
paramount importance if future surgery is to be successful at creating muscle cineplasty interfaces.
Muscle tone can be preserved by ensuring that residual muscles retain the ability to develop tension
when voluntarily contracted. This requires that either myoplasty and/or myodesis be performed at the
time of initial amputation. In myoplasty, agonist-antagonist residual muscle pairs are tied off against
each other. In myodesis, the residual muscle is stitched to the bone. In both cases, the residual
musculature retains the ability to easily develop tension, thus preventing atrophy during the interval
between initial amputation and the revisions necessary to create a muscle/tendon cineplasty control
interface. Myoplasty would generally be used on the superficial muscles, whereas myodesis could be
used on the deep muscles.
In the area of functional electrical stimulation, skeletal muscle transfers (myoplasty) combined with electrical stimulation have been advocated to provide contractile function that augments or replaces impaired organ function (Grandjean et al., 1996). Clinical investigation of dynamic cardiomyoplasty for the treatment of heart failure and dynamic myoplasty for treatment of fecal or urinary incontinence is already under investigation.

In Sweden, pioneering work in the area of direct skeletal attachment has been performed by Brånemark and his team (Brånemark, 1997; Brånemark et al., 2001) The technique is called osseointegration, and these surgeons and orthopedic engineers have created interfaces for direct skeletal attachment systems for upper and lower limb amputations. With osseointegration, a prosthesis is attached directly to the skeleton by way of a titanium abutment that protrudes through the skin from the cut end of the bone. Brånemark’s techniques appear to have greatly diminished the infection problem that persisted in previous efforts (Hall and Rostoker, 1980). Should direct skeletal attachment prove itself viable, it could revolutionize the prosthetic fitting of some amputation levels.

The work of Kuiken (1995) is another example of the trend toward innovative surgical procedures. He advocates the use of neuromuscular reorganization to improve the control of artificial arms. Kuiken observed that although the limb is lost in an amputation, the control signals to the limb remain in the residual peripheral nerves of the amputated limb. The potential exists to tap into these lost control signals using nerve-muscle grafts. As first suggested by Hoffer and Loeb (1980), it may be possible to denervate expendable regions of muscle in or near an amputated limb and graft the residual peripheral nerve stumps to these muscles. The peripheral nerves would then reinnervate the muscles, and these nerve-muscle grafts would provide additional control signals for an externally powered prosthesis.

The nerve-muscle grafts could potentially be directly attached to force sensors using microcineplasty techniques and provide extended physiologic proprioception (EPP). Alternatively, the surface EMG from the nerve-muscle grafts could serve as a traditional myoelectric control signal, using existing myoelectric technology. These grafts could provide simultaneous control of at least the terminal device and powered elbow and possibly another degree of freedom such as a wrist rotation or wrist flexion-extension. In a shoulder disarticulation amputee, each of the residual brachial plexus nerves could be grafted to different regions of the pectoralis muscle. The prosthesis of the future might be a multifunctional device mounted into the skeleton using osseointegration and controlled using multiple miniature muscle cineplasties.

### 32.8 IN CONCLUSION

To put the current state of prosthesis control in the context of the evolution of control in airplanes, automobiles, and remote manipulators, it can be seen that all these fields used similar control
methodologies in their early days. However, prosthetics later diverged from the others with the advent of electric power and myoelectric control. In each field, flight surfaces, front wheels, remote manipulators, or prosthetic joints were controlled directly by linking the operator to the device. In early aircraft, control cables connected the joystick to the wing, tail flaps, and rudder. In early automobiles, a rack and pinion mechanism connected the driver directly to the front wheels. In early master-slave manipulators, master and slave were physically connected by cables. In early prostheses, a control cable connected the user to the artificial joint.

In each of these cases, the operator provides both power and control signals and transfers them to the machine via a mechanical linkage that physically interconnects the operator and the machine output. Feedback from the machine is transferred back to the operator via the same mechanical linkage, i.e., feedback is returned to the user via the control interface (Fig. 32.33). The pilot could “feel” what was happening to the aircraft flight surfaces through the joystick. The amputee had a sense of prosthesis state through the control cable.

As aircraft got larger and faster, hydraulic power assist was added to the cable controls. As automobiles advanced, power steering was developed. The control input (the operator at a joystick or steering wheel) remained the same, but instead of the operator providing both the power and control signals, power was now provided by hydraulic systems, which augmented the cable systems of aircraft and the steering of cars. The key is that the same type of feedback through the control linkage was maintained. Position, velocity, and force information was still fed back with these “boosted” systems. A similar pattern of development occurred with remote manipulators (i.e., human-powered cable systems were followed by electric motor boosted cable systems).

However, it was at this point, the advent of externally powered devices, that prosthetics diverged from the others and moved almost exclusively to open-loop velocity control techniques with its associated loss of feedback information. It is the lack of subconscious sensory feedback to the user that has inhibited the use of many externally powered multifunctional prosthetic designs.

Meanwhile, in the area of remote manipulation, electrohydraulic, or electromechanical, bilateral master-slave systems were developed for the nuclear power industry to reduce the mental burden placed on the operator by providing force reflection. Aircraft controls for large planes and some high-performance military aircraft, building on work from the field of teleoperation or remote manipulation, have done away with the direct physical interconnection of the pilot and flight surfaces and now use fly-by-wire systems.

Fly-by-wire systems owe much of their development to Goertz (1951) who built the first bilateral force reflecting master-slave remote manipulators, and Mosher (1967) at General Electric Corp, who built force reflecting manipulators with power augmentation. In these systems, the pilot/operator is connected to the flight surfaces through electrical wire connections, and the systems use bilateral master/slave techniques with force feedback or force reflection to provide the operator with the appropriate feedback or “feel.” Automobiles are also beginning to go to steer-by-wire and brake-by-wire systems (DesignFax, 2001).

For the top-of-the-line fighter planes, a stage beyond fly by wire has been reached. These planes are so inherently unstable that a human operator is incapable of controlling them unaided, so now boosted control has been added to boosted power. Artificial reflex-based systems are the logical conclusion of this trend to move more and more of the control from the operator to automated systems. In the end, the operator is just along for the ride. In fact, operatorless systems are being explored in a new generation of military aircraft and were employed in the U.S. action against Afghanistan in 2001. These aircraft are essentially robot planes that are piloted from the ground. For these aircraft, performance is no longer limited by the bandwidth of the pilot or by what the pilot can withstand in terms of “g” forces.

As for arm prosthetics, the future of what can be achieved depends to a large extent on the interactions of physicians, surgeons, prosthetists, and engineers. If meaningful multifunctional control of prostheses is to be achieved then physicians and surgeons need to perform innovative procedures that can be coupled with the novel components and controllers that engineers and prosthetists create.
ACKNOWLEDGMENT

The author is deeply indebted to Dr. Dudley Childress and Mr. Craig Heckathorne of the Northwestern University Prosthetics Research Laboratory, Chicago, Illinois, for taking the time to review this document.

REFERENCES


Childress, D. S. (1992). Control of limb prostheses, Chapter 6D In *Atlas of Limb Prosthetics, Surgical, Prosthetic, and


Lee Publications (2001). Gung Ho Grip™ A-Okay, *Lee’s Action Figure News & Toy Review (AFN&TR)*, Lee Publications, Monroe, Conn. no. 102, p. 16, April.


