DESIGN OF A POWERED PROSTHETIC ARM SYSTEM FOR THE ABOVE-ELBOW AMPUTEE[a]

Carl Peter Mason, M.S., B.E.
Senior Engineer

Bioengineering Research Service
Veterans Administration Prosthetics Center
252 Seventh Avenue, New York, N.Y. 10001

INTRODUCTION

After World War II, a giant step forward was taken in American upper-extremity prosthetics. Sockets of laminated plastic, molded over plaster-of-paris casts of a man's stump, were made to replace those that were hand carved and hand worked. The Bowden cable and the split-finger hook were accepted as active prosthetic components, and the locking mechanical elbow was invented. The level of functional replacement of these prosthetic devices was significantly higher than the devices available before this era.

Since then only minor improvements have been made which have actually reached the average patient. Now, upper-extremity prosthetics is again ready to take another giant step forward to significantly improve the performance of the average amputee. This is now possible due to advances that have taken place in the state of the art of many related fields.

Miniature electronic circuits, utilizing both integrated and hybrid techniques; economically reasonable high-performance miniature motors; advances in materials science, molding techniques, the state of the art in batteries or electrical energy storage; and better understanding of the basic myoelectric characteristics of an amputee's remaining stump muscles—all contribute to the development of new prostheses.

New fabrication techniques permit construction of a total-contact, self-supporting socket. Sockets utilizing this intimate technique and myoelectric signals permit the total self-containment of an external power source within a prosthesis.

---

[a] Based on a thesis submitted to the Polytechnic Institute of Brooklyn, N.Y., in partial fulfillment of the requirements for the degree of Master of Science in Bioengineering.

It is the purpose of this article to show that a prosthesis, capable of improving the performance of the average amputee, can be designed utilizing the synergistic effect of all of these engineering advancements.

FUNCTIONAL PERFORMANCE OF NORMAL MAN DISTAL TO THE SHOULDER

Studies (1,2) conducted on normal motion, indicate that the most common motion is elbow flexion (16.5 percent of all motion). The second most common action is hand prehension as a general category (14.1 percent). Upper-arm internal rotation and forward flexion accounted for 11.9 percent and 11.0 percent, respectively. Wrist flexion and forearm pronation each accounted for 9.5 percent. Using one elbow flexion as the standard unit of measure and comparing the range of motion, percent used most frequently, average power for the most frequently used part of motion, energy expenditure during the most frequently used part of motion, the maximum necessary output torque, and the average power and total energy, we can approximate the requirements for the activities of daily living. The part of the maximum range used most frequently in elbow flexion is from 90 to 130 deg. and this is used 60 to 90 percent of the time. The average power is 40 ft.-

TABLE 1.—*Average Power for Normal Healthy Subjects Manipulating Weights (6–21 lb.).*

Motion	Relative freq.	Ave. power ft.-lb./sec.	Energy of most freq. used motion, ft.-lb.	Max. torque in.-lb.
UPPER ARM				
Forward flexion	.667	50	7.8	300
Extension	.248	30	1.2	300
Abduction	.563	70	5.1	380
Internal rotation	.720	80	3.8	280
External rotation	.061	75	3.7	230
FOREARM				
Flexion	1.000	40	9.0	100
Pronation	.575	13	5.5	80
Supination	.200	11	4.4	85
WRIST				
Flexion	.575	8	2.2	60
Extension	.194	8	2.2	60
Adduction	.302	6	1.6	40
Abduction	.097	5.8	1.8	40
HAND PREHENSION	.855	9 max.	5.5 max.	40

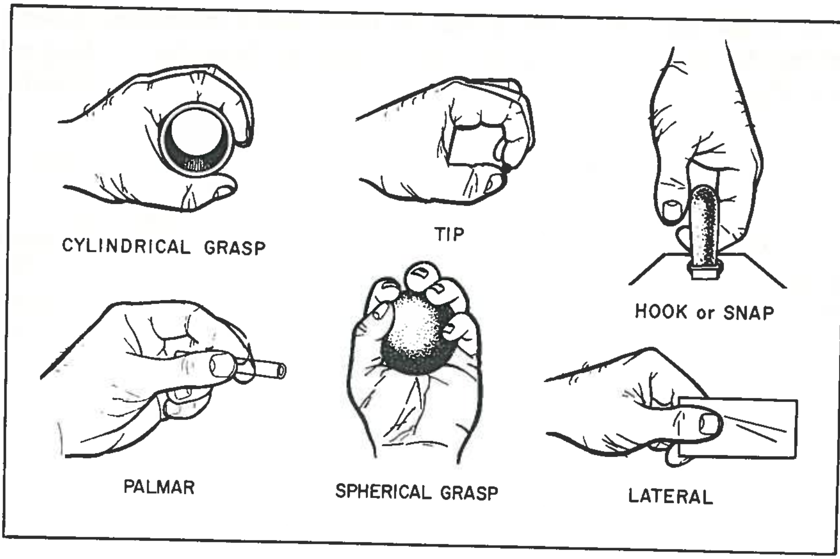

FIGURE 1.—The various types of hand prehension.

lb./sec. The energy expended during this most frequently used motion is 9 lb.

Hand prehension has a correlation of .855 for each elbow motion and is divided into six different types of prehension: palmar, lateral, cylindrical, spherical, tip, and hook (Fig. 1). In picking up objects, prehension is divided among the above in a ratio of 40, 24, 21, 6, and 9 percent, respectively. In holding or retaining an object for use or transportation, palmar prehension is completely dominant, being used 81 percent of the time (as when writing or utilizing hand tools). Lateral is used 8 percent, cylindrical 6 percent, and hook 5 percent. The maximum power for hand prehension is 9 ft.-lb./sec. Maximum energy is 5½ ft.-lb. and maximum torque is 40 in.-lb. A complete breakdown of torques and powers may be seen in Table 1.

DESIGN REQUIREMENTS IN AN ABOVE-ELBOW PROSTHESIS

Minimally, a prosthetic replacement should match the kinematic and kinetic capabilities of the limb segments that they are replacing. Until a method of directly communicating with the afferent nervous system is perfected, complete duplication of a lost limb segment will never be possible. Even the most sophisticated laboratory experimental devices are presently only generating information as to prosthetic hand prehension and position (Hannes-Schmidl and INAIL), or elbow position and torque (Boston Elbow, Liberty Mutual). The other information we gather through our limbs is impossible to duplicate. We can feel the shape, texture, temperature, and dynamics of objects without seeing them. This is not feasible in prosthetics in the foreseeable future. Likewise, the independent control we have of simultaneously and automatically moving our hands and arms in three-dimensional space will not be achievable until a method of tapping directly into the efferent nervous system is realized.

Unfortunately, due to the limitations imposed in supporting a prosthesis with a socket instead of with direct bone attachment, some very severe limitations are recognized. The weight of your arm is not normally even sensed, but if this arm were supported by an artificial socket and harness transmitting its load through the skin, this load would act as a continuous irritant by the end of the day. It would feel many times heavier than it actually is.

Presently available hooks weigh between 3 and 13 oz. The typical weight is approximately 7 oz. The typical weight for a forearm and wrist unit (9 in.) as commercially procured is 11 oz. The weight of the Hosmer Internal Elbow varies from 12¼ oz. (for the E-200) to 14¾ oz. (for the E-400). The typical weight for a mid-level above-elbow socket is about 1¼ lb., and a Figure-8 harness adds approximately 2 oz., bringing the weight of a typical well-made prosthesis to approximately 52 oz., or 3 lb. 6 oz. Realistically, if one were to power a prosthesis externally, the total weight should not exceed the present devices.

Dimensionally, prosthetic components are usually smaller than their anthropomorphic counterparts. To allow for fitting the greatest number of amputees, the prosthetic components are usually dimensionally designed to extend only the minimum distance above the replaced joint.

The power requirements for a prosthetic system may be determined (2) by combining the typical number of times an amputee will flex his elbow (300 average, as determined by a counter on a patient's elbow) with the energy requirement for a non-amputee to perform a typical activity of daily living. A realistic value for mechanical energy output for an above-elbow patient would be 7100 ft.-lb., or 2.7 watt-hours.

RELATING MYOELECTRIC SIGNALS TO PROSTHETIC FUNCTIONS

Basic myoelectric (EMG) signals are detectable at the surface of the skin in amplitudes ranging from 10 microvolts through 5 millivolts (r.m.s.). The frequency spectrum is centered between 100 Hz and 150 Hz, with a rapid low-frequency roll-off and a longer high-frequency range (Fig. 2). These EMG signals have been used in a number of ways to control prosthetic arm components.

In the earlier systems, rectified signals from two antagonistic muscles were detected and compared to determine the stronger one which activated a relay to make a device move in a particular direction. This is

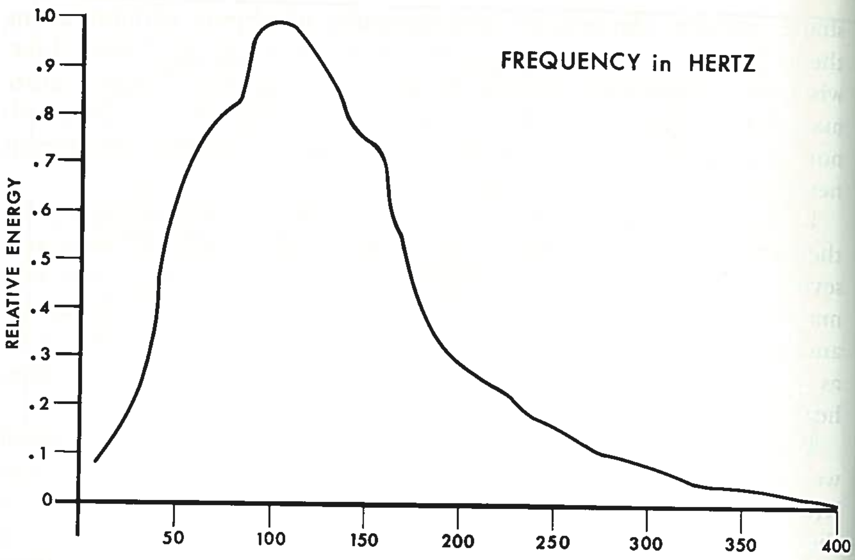

FIGURE 2.—Typical myoelectric signal spectrum from upper-extremity skeletal muscle.

generally called *digital control,* in which one muscle makes a component perform one action when its signal exceeds a threshold value, and signals from the antagonistic muscle make the device perform the opposite function when its actuation threshold is exceeded. Systems such as this are sometimes called *two-muscle site/two-control digital* (on-off) systems. The early Russian, Bock, and Viennatone hands function in this manner.

A more physiologically natural system can be built if the raw signals are used to control not only on-off conditions, but also the *rate* of function of the prosthesis. The newer Russian, Viennatone, and Veterans Administration and Northwestern University (3) systems operate on this principle. This is sometimes called a *two-site, two-function, proportional control system.* The Boston elbow also operates in this manner, but in addition it has internal feedback to insure that speed and torque are related to the muscle signals in a linear manner.

A modification of this system completely compatible with conventional prostheses is the *single-site, single-function, proportional control system.* In these devices a signal from a single muscle is used to drive a prosthetic device in one direction with a speed or force proportional to the signal. The opposite function is obtained by gravity or springs. The Johns Hopkins (4) externally powered system operates in this manner.

As the level of amputation rises, the number of useful muscle control sites decreases. This makes it necessary to utilize a single site to control two or more functions (*single site, dual control*). EMG signals may be interpreted in various ways to achieve this. The r.m.s. amplitude of the signal may be divided between the repeatedly achievable minimal and maximal level, into discrete function/no-function levels. Divided, the signal actually electronically creates several threshold levels. The early control systems developed by R. N. Scott (5) operated in this manner. A mild muscle contraction generating small EMG signals produced one function. At a higher level the opposite function occurred. Bousso, in Israel, also utilized this type of EMG control to drive gas-solenoid valves to control a gas-powered prosthesis. A more advanced form of this type of control is presently being used by INAIL and Scott. In this single-site, dual-control, proportional system, a minimal to sub-maximal signal is used to drive the prosthetic component in a proportional manner. A maximal contraction (exceeding the upper threshold) is used to drive the prosthesis in the opposite direction at a fixed rate (digitally).

A third method of utilizing a single site to control two functions in a semi-proportional manner is being used by Northwestern University and the Veterans Administration. In this technique, the *rate* of muscle contraction controls the prosthesis in a proportional manner. If a muscle signal is generated gradually, the device responds proportionally to the amplitude in one direction. A muscle signal generated by a rapid contraction causes the prosthesis to operate in the opposite manner at a fixed digital rate.

A fourth method is to utilize a single muscle site to proportionally drive a prosthetic device in *both* directions. From one viewpont, this may be considered a physiological analog in that a strong contraction of a muscle actuates one function against not only gravity, but also the lengthening of the antagonistic muscles. Total relaxation of this muscle results in the opposite function. This may be achieved by using a very small signal to make the device operate in one direction at its maximum rate. As the signal increases, the device operates in the same direction but at a proportionally reduced rate. When the muscle signal is maintained at a *low-to-moderate* rate, the device stops. As the muscle signal increases above this range, the device is driven in the opposite direction at an increasing rate related to the increasing muscle signal. This method is utilized by the Veterans Administration (6) to control an elbow.

Other methods of controlling the prosthesis from an EMG signal may be envisioned but those described above have already been tested on patients.

ELECTRICAL ENERGY STORAGE SYSTEMS

Presently, there are seven commercially available types of portable electrical energy storage. Each has its advantages and disadvantges for use in a prosthetic device. Ideally, we would like the system to have a cycle life in excess of a year, to have a minimum weight, minimum volume, and maximum energy storage commensurate with the rest of the prosthetic system. In addition, it should be in no way dangerous and, in the event of failure, it should *gradually* malfunction before going completely dead.

The rechargeable alkaline-manganese battery has a positive electrode of MnO_2, a negative electrode of zinc (Zn) and utilizes KOH (ZnO) as the electrolyte to achieve 1½ volts per cell. These batteries are presently used in tape recorders, strobes, and portable TV sets. Their usefulness is limited by a cycle life of between 30 and 50 cycles, and they cannot deliver large current discharges, which makes them unsuitable for prosthetic applications.

Lead-acid batteries, utilizing PbO_2 as the positive electrode, H_2SO_4 for the electrolyte, and Pb for the negative electrode, generate 2 volts per cell and have a full charge-discharge life between 100 and 400 cycles. These cells cannot be used unless they are sealed to protect against acid spillage.

Sealed lead-acid batteries operate on the same principles as their unsealed counterpart with a slightly reduced cycle life. They can operate in any position safely and are presently used in portable TV sets and portable appliances. They deserve further consideration for prosthetic applications.

Nickel-cadmium batteries utilizing NiOOH for the positive electrode, KOH for the electrolyte, and cadmium for the negative electrode, generate 1.2 volts per cell and have the longest cycle life of any storage system (betwen 500 and 5000 charge cycles). They are presently used in standby power supplies and power-distribution boards. Because they contain a corrosive liquid, KOH electrolyte, they must be used in the upright position; these two conditions make them unsuitable for prosthetic applications.

Sealed nickel-cadmium batteries overcome the above difficulties, paying the penalty of a slightly reduced cycle life (200 to 1000 cycles) and delivering even larger current discharges. They utilize the same electrochemical reaction as the unsealed versions. These batteries are very popular in portable electrical accessories and are very promising for prosthetic applications.

Silver-zinc cells have high theoretical energy density, more than twice the theoretical value for a nickel-cadmium cell. They can deliver 1½ volts per cell and very large current discharges. Unfortunately, those cells with long life have low high-current discharge capabilities and those cells with high-current capabilities have very limited life. The range of charge cycles varies from 20 to 400 cycles, which makes the presently available cells unsuitable for prosthetic applications.

Silver-cadmium batteries are similar to silver-zinc but they use cadmium as the negative electrode, generate 1.1 volts per cell, have a large current discharge capability, and a cycle life between 300 and 2000 cycles. This is the longest cycle life of any sealed cell usable in prosthetics. They are presently used in communication equipment, in aircraft, and they deserve further consideration for prosthetic systems.

The sealed lead-acid battery which comes closest to the requirements for incorporation in a prosthetic device is the Globe "Gel-Cel" No. GC 610-1. This battery costs $6.85 each (OEM price, Globe Battery Division, Globe-Union, Inc., 5757 N. Green Bay Ave., Milwaukee, Wisc. 53201). This battery provides 6 volts and 900 mah. (at the 20-hr. rate), weighs 8.8 oz., and stores 9.8 watt-hours per lb. This energy-delivery capability is significantly reduced when the cell is discharged at the 1-hour rate. It will last approximately 8 minutes when discharged to 4.8 volts at the 1-hour rate (.9 amp.).

The Silvercel (Yardney, No. 8-HR-05, Yardney Electric Corp., 82 Mechanic St., Pawcatuck, Conn. 02891) has extremely high theoretical energy storage. Unfortunately, a ½-ah. 12-volt battery, weighing almost 7 oz. and occupying just under 8 cu. in., can only be recycled 15 to 20 times when it is used at the 1-hour rate, which makes this battery impractical for prosthetic applications.

The sil-cad cell, similar in construction to the Yardney silver-cell, has a significantly longer cycle life. A 10-YS-05 battery (Yardney) has a nominal capacity of ½ ah. and an initial voltage of 11 volts. This battery weighs approximately 8 oz. and occupies approximately 10 cu. in. There is a 30 percent reduction in the storage efficiency when the battery is discharged at the 1-hour rate, and if the battery is to be cycled over 300 cycles, it cannot be discharged to more than 70 percent of its capacity. These two factors cut the effective capacity approximately in half.

The Eveready nickel-cadmium battery Y6130A (Union Carbide Corp., Consumer Products Division, 270 Park Ave., New York, N.Y. 10017), is composed of 10 CF450 cells and has a nominal capacity of 450 mah., weighs less than 7½ oz., and occupies 4 cu. in. This cell may be charged at the 1-hour charge rate (with proper electronic controls) faster than any cell so far mentioned. It can deliver 450 mah. at the 1-hour rate if discharged to 10 volts and is derated only 8 percent if a conservative 1.1 volt cell cutoff is used. This battery can be charged over 500 times if it is discharged to over 95 percent of its capacity, and a reduction of only 20 percent in its nominal capacity is considered as the end of its recharge life. This cell effectively stores over 5 watt-hours of energy and can meet all of the other requirements of a prosthetic system.

In summary, even though nickel-cadmium batteries offer the lowest theoretical energy storage, because of their advanced state of the art, they offer the highest energy storage ratings in the sizes of interest to the designer of an electrically powered prosthetic system.

MOTOR DESIGN

With the advent of transistor servo technology, capable of extreme miniaturization, fast response, high efficiency, and high power, the need for motors with matching characteristics was generated. Conventional motors either have a wound field or a permanent magnet (PM) field surrounding an iron armature. The PM type is generally more efficient because no power is wasted generating a magnetic field for the armature to react with. A conventional iron armature has a relatively large inductive mass. This, combined with the self-inductance of the windings, tends to give conventional motors significant inertia and self-inductance, both of which tend to limit the motors' starting time and response.

A motor with an ironless armature and as much copper as possible within the air gap of the magnetic system solves this problem.

Presently, there are three commercially available methods of achieving this. A printed-circuit motor with a flat printed-circuit disk-shaped armature and radially located PM's surrounding this disk can achieve high torques and relatively fast responses. Unfortunately, these units are not available in the size needed for upper-extremity prostheses.

A bell-shaped armature has the advantage of very low inertia and relatively short starting times. This type armature has the permanent magnet in the core location and an iron-flux return surrounding the outside. This generates a high uniform magnetic field for the armature to react with. The only additional mass in the system is in the armature support and the coil-winding heads. An even faster starting, more efficient armature can be built by skew winding of the armature (Fig. 3). This has the additional advantage of lowering the ohmic resistance by eliminating the coil-winding head and connectors and by making the armature self-supporting because the wires pass perpendicular to one another. Starting time constants of 10 milliseconds are achievable. Miniature commutators usually made of precious metal combinations permit extremely low electrical contact resistance and low mechanical friction. Motors with this configuration are capable

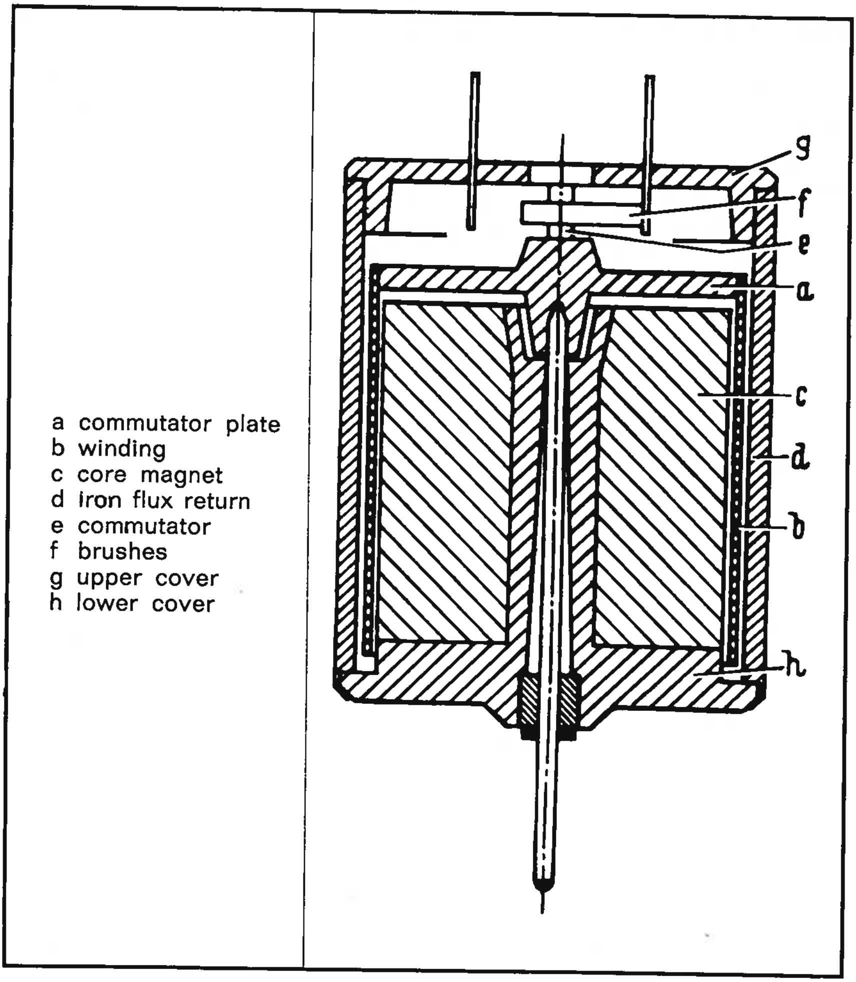

FIGURE 3.—Sectional view of a DC-Micromotor with skew-wound bell-type armature.

of operating at electrical to mechanical efficiencies approaching 90 percent and are capable of delivering their maximum power at almost 50 percent. These characteristics of high efficiency, low inertia, fast starting times, and moderate cost make them ideal for prosthetic applications.

PLASTIC MATERIALS

Presently, there are approximately 30 different basic synthetic organic resin plastics. These are either thermoplastic or thermosetting resins; each of these may come in a variety of forms, and each form may be combined with another multi-dimensional array of alloying materials. In prosthetic applications, where one wishes to take advantage of the higher structural properties of plastic, a smaller choice of materials is available.

Nylon is commercially available in five different types with or without reinforcement or filling. Forty percent long-fiber Fiberglas-type 6/6 nylon (7) has a tensile strength of 30,000 p.s.i., a flexural modulus of 1.8×10^6 p.s.i., water absorption of 0.6 percent in 24 hours (0.0002 in./in.), and a coefficient of thermal expansion of 1.1×10^{-5} per degree Fahrenheit.

Twenty percent Fiberglas acetal copolymer resins (7) have a tensile strength of 18,500 p.s.i., a flexural modulus of 1.1×10^6 p.s.i., water absorption of 0.29 percent, and a coefficient of thermal expansion of 1.9×10^{-5}. The acetal resins have a low coefficient of friction against other materials, 0.15 against steel.

Forty percent glass-reinforced polycarbonates (7) have a tensile strength of 21,500 p.s.i., a tensile modulus of 1.7×10^6, water absorption of 0.09 percent in 24 hours, and a coefficient of thermal expansion of 1.2×10^{-5}. This combination of properties makes this material ideal for the economic production of prosthetic components that require tolerances unachievable with earlier plastic materials.

THE DESIGN OF THE SYSTEM

To successfully design a prosthesis giving an amputee improved function, all of the other criteria presented must be met. Efficiency in the full meaning of the word is the key. The limitations of the conventional system must be overcome and new limitations not created. The weight of the entire system must be less than the conventional system. It must have less harnessing, have better control, have more comfort, be more economical, be cosmetically acceptable, be reliable, and operate in a natural manner.

This can be achieved by improving all of the components and dividing the weight differently than in the conventional system. When a patient senses the weight of the prosthesis, he feels not only the magnitude of the inanimate components, but also the dynamics as reflected in the center of gravities of the various components and the restrictions of the socket and the harness, as they too generate the sensation of weight. Some weight reduction can be accomplished in every subcomponent of the system except, possibly, the hand, and these reductions allow the external energy source to be incorporated within the devices. If electrical-to-mechanical efficiency is maintained at the highest possible levels and the electrical control efficiency is also extremely high, the requirements for energy storage in a high-efficiency storage system will not exceed the limitations imposed by the weight of the conventional system.

The conventional mechanical hand weighs 13 oz. and, with the exception of two voluntary closing hands, generates 2½ to 7½ lb. prehension force. Before designing a new powered hand, the only two commercially available units were investigated, but both had limitations.

One hand achieved high prehension force and speed by using an automatic load-sensing planetary gear shift to increase the motor speed reduction system when the hand met resistance. The planetary system was sealed for silence. It was controlled in a digital manner and the automatic transmission made it difficult to incorporate proportional control. The sophistication of the transmission reduced the system's reliability and the use of an external battery pack increased the harnessing on the patient.

The other hand also had digital control and an external battery pack. This hand utilized a four-stage spur-pinion reduction system and a conventional PM motor. This hand was used as the basis for the VAPC design because it was adaptable to proportional control and for the production economy in using a commercial hand frame. The unit had a one-way mechanical clutch that permitted the motor to drive the hand closed but stopped the load on the hand from back-driving the motor. This also permitted the gear train to be operated at a high efficiency.

To improve the operating characteristics and reduce the noise, the motor and the first two stages of gear reduction were replaced. A special skew-bell wound motor was designed for reduced speed, increased torque, and increased life characteristics. The metal spur-pinion reduction gearing was replaced by helical plastic gears with a higher first-stage reduction ratio and a lower second-stage reduction ratio. The finger assemblies were modified to incorporate a finger breakaway to release an amputee's grip in an emergency. The finger and thumb tips were replaced with shaped pads to improve the prehension characteristics. The electronics (designed by Dr. Dudley S. Childress, Northwestern University) and battery pack for the below-elbow amputee were incorporated in the wrist. An oval-to-round transition section with a built-in quick-connect and rotational friction was designed. Plain stainless-steel electrodes were used in a configuration designed by Northwestern University to mate their self-suspended socket.

The conventional forearm is a plastic laminated shell with a wrist friction unit on the distal end and an elbow saddle on the proximal end. The 11-oz. forearm was replaced by the endoskeletal urethane foam-covered forearm consisting of a reduced-weight wrist friction unit, skeletal aluminum tube, and an aluminum elbow saddle. The new unit was soft to the touch, did not make noise when hit against environmental objects, and weighed only 7 oz., 4 oz. less than the standard.

The conventional mechanical elbow weighs over 12 oz. There were no commercially available adult powered elbows. To design a powered elbow utilizing conventional gears and weighing less than the conventional elbow is almost impossible. To achieve the impossible, a nonconventional system was designed. A harmonic drive was designed to provide a high reduction ratio and still be efficient and light. To achieve this high strength, plastics had to be injection molded to tight tolerances, and the material had to keep these tolerances. The housing, bearings, control-switch mountings, and rigid spline of the harmonic drive were designed into one-half of a mirror-image injection-moldable assembly. Twenty percent fiber glass 15 percent FTE-reinforced polycarbonate was used for this piece. The flexspline, the motor support, thermal heat sink, and limit actuators were designed into a second injection-moldable piece. Twenty-five percent long-fiber acetal resin was used for its high strength and low friction, necessary characteristics in this part. A hardened tool steel-roller planetary wave generator was designed to take advantage of the relation between radial forces and output load on the wave generator. The motor shaft was also sleeved with tool steel to act as the sun roller of the planetary wave generator and to support concentrically the entire wave generator body (Fig. 4). The motor was specially designed to take the necessary load using alnico V for the core and a skew-bell wound armature with a special brush configuration. Electrical limit switches were used in a diode blocking matrix to limit the travel of the motor and to stop the motor from being stalled in the end position. This permitted the motor to be operated at twice its rated voltage, delivering four times its rated power for a short-duty cycle.

The overall system efficiency was adjustable by varying the sun diameter a few thousandths of an inch, thereby increasing the preload on the wave generator and flexspline. This system was set at the maximum efficiency (approximately 50 percent) that permitted self-locking when the motor was shunted either by the limits or the control electronics or control switches. The elbow was attached to the humeral section by either a turntable matching the standard humeral section, permitting retrofitting in a conventional arm, or by a quick-connect turntable that provided frictional control and space for the batteries.

The battery pack was designed to utilize the latest fast-charge high-energy nickel-cadmium cells sufficient for both hand and elbow opera-

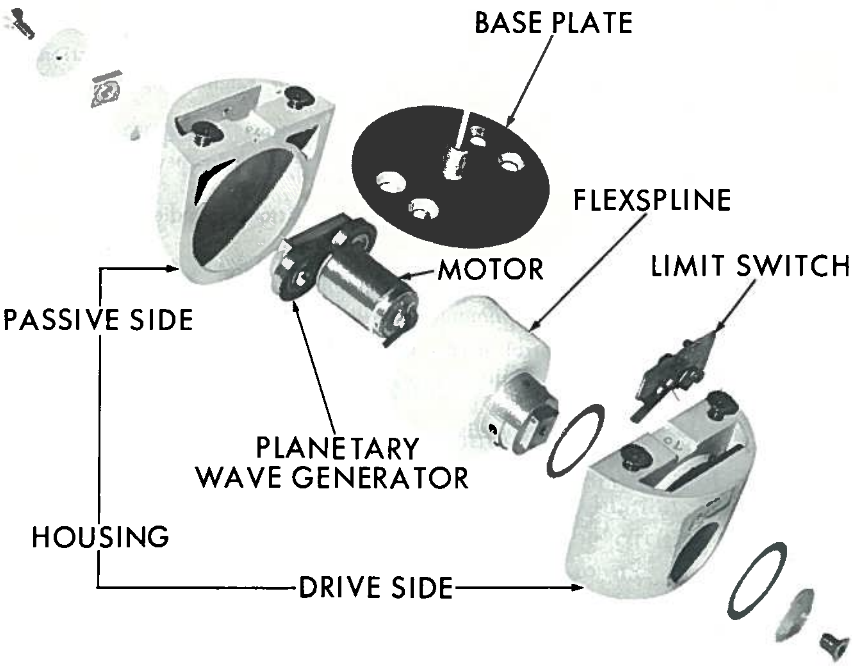

FIGURE 4.—Exploded view of VAPC Electric Elbow.

tion. A thermal hold circuit was designed to permit the cells to be charged at the 1-hour rate and to automatically switch to a 40-hour trickle when fully charged. A ring to mate the elbow quick-connect was designed to be laminated into the humeral section of the prosthesis. Acrylic lamination utilizing a dual vacuum system was suggested to permit the fabrication of a lighter socket.

SUMMARY AND CONCLUSIONS

All of the aforementioned designs and features, when used independently or in total, can create an improved prosthetic arm system for the above- and below-elbow amputee. This is a system that can offer reduced harnessing, increased cosmesis, increased comfort, and reduced weight at an economical price.

There is still a great deal to be done to increase the information transfer between the prosthesis and the amputee. If the field continues moving at its rapidly accelerating rate, this should be realizable in the near future.

REFERENCES

1. Keller, A. D., C. L. Taylor, and V. Zahm: Studies to Determine the Functional Requirements for Hand and Arm Prosthesis. Dept. of Engineering, University of California at Los Angeles, 1947.
2. Paul, I. and R. W. Mann: Evaluation of Energy and Power Requirements For Externally Powered Upper Extremity Prosthetic and Orthopedic Devices. A.S.M.E. Paper # 62-WA-121, July 1962.
3. Childress, D. S.: Design of a Myoelectric Signal Conditioner. Audio Engng. Soc., 17:286–291, June 1969.
4. Seamone, W. and G. Schmeisser: Development and Evaluation of Externally Powered Upper Limb Prosthesis. Bull. Prosthetics Res., BPR 10–16:169–176, Fall 1971.
5. Scott, R. N.: Myo-Electric Control. Science J., 3:53–59, 1966.
6. Research Report, Veterans Administration Prosthetics Center. Bull. Prosthetics Res., BPR 10–15:209–211, Spring 1971.
7. 1971 Plastics/Elastomers Reference Issue. Machine Design, Volume 43, February 1971.