June 1981
Volume 35
Number 2

Orthotics and Prosthetics

Journal of the American Orthotic and Prosthetic Association

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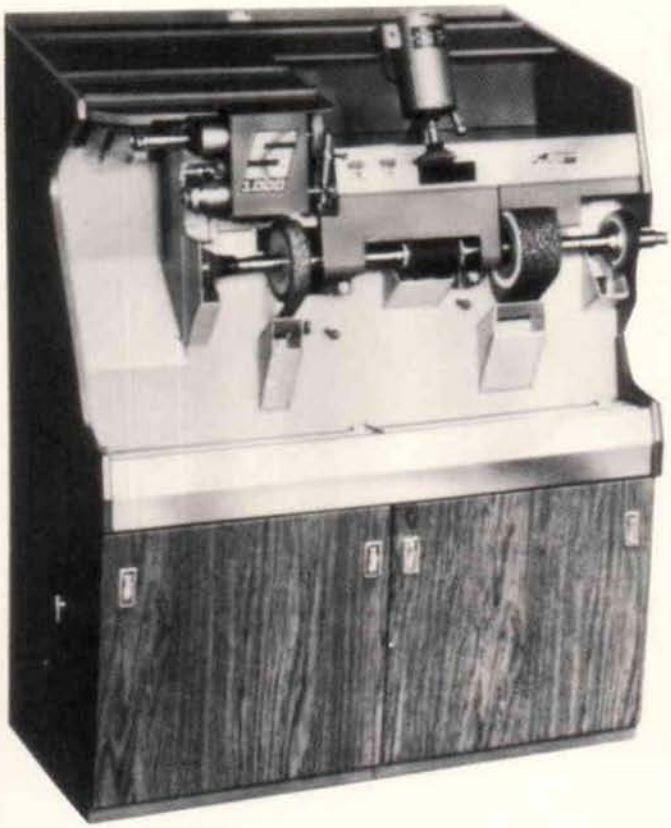

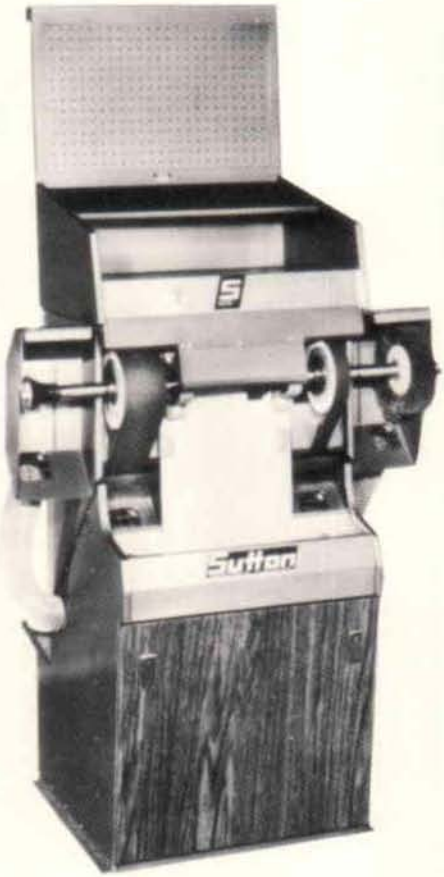

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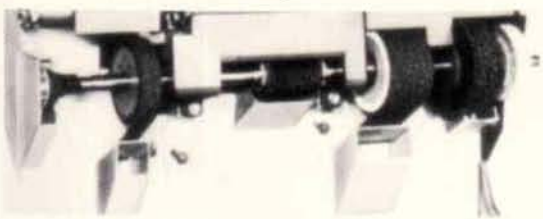

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Orthotics and Prosthetics

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Meetings and Events

1981, August 24-27, University of New Brunswick, Frederiction, New Brunswick, Canada. Course: Myoelectric Control of Artificial Limbs. Contact: Director, Myoelectric Controls Course, Bio-Engineering Institute, University of New Brunswick.

1981, July 24-25, AAOP Seminar, Atlanta Marriott Motel Hotel, Atlanta, Georgia.

1981, September 9-11, First Annual Advanced Course in Lower Extremity Prosthesis, Nassau County Medical Center, East Meadow, New York

1981, September 18-19, AAOP Workshop, Houston, Texas.

1981 October 27-November 1, AOPA National Assembly Sahara Hotel, Las Vegas, Nevada.

1981 November 21-22, AAOP Seminar, California.

1981 December 9-12, AAOS Seminar Sheraton, Miami Beach, Florida

1981 December 13, AAOS Workshop Sheraton, Miami Beach, Florida

1982 February 17-20, AAOP Annual Meeting and Round-up Seminar, Royal Sonesta Hotel, New Orleans, Louisiana.

1982 April 16-17, AOPA Region I Meeting, Marriott Hotel, Worcester, Massachusetts.

1982, April 29-May 2, AOPA Regions VII and VIII Combined Meeting, Alamada Plaza, Kansas City (Tentative).

1982, May 6-9, AOPA Region IV Meeting, Radisson Plaza Hotel, Mashville, Tennessee.

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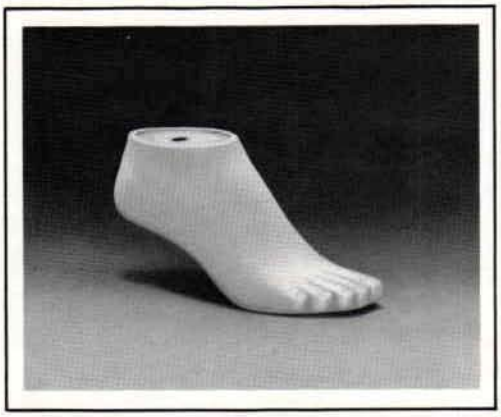

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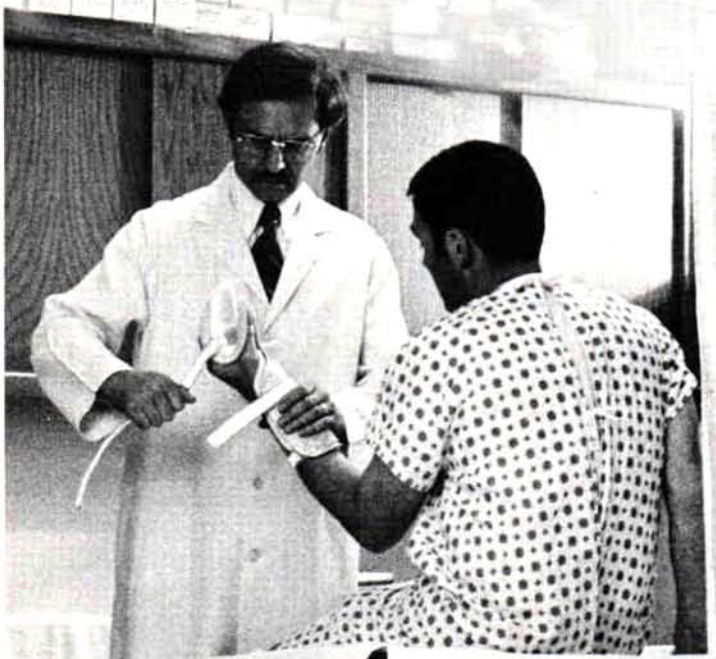

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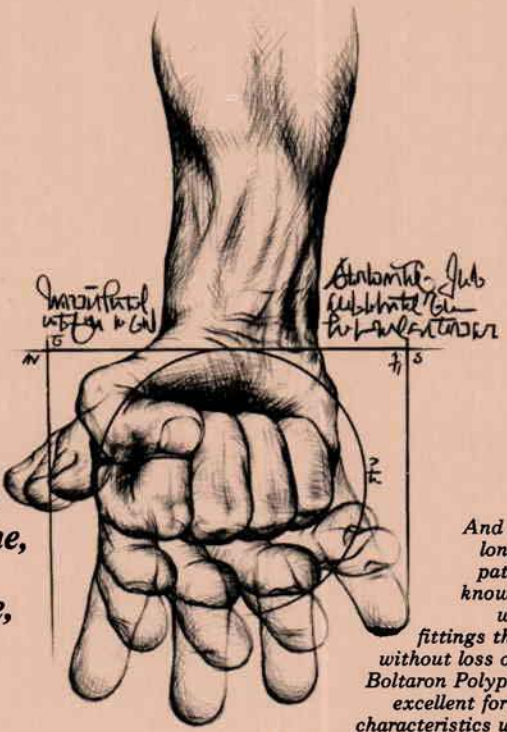

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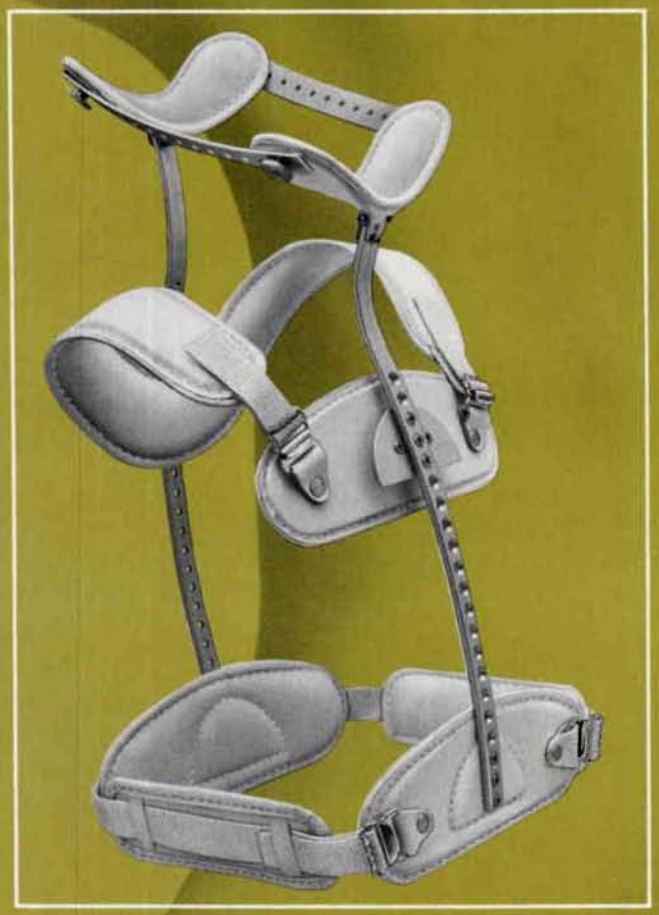

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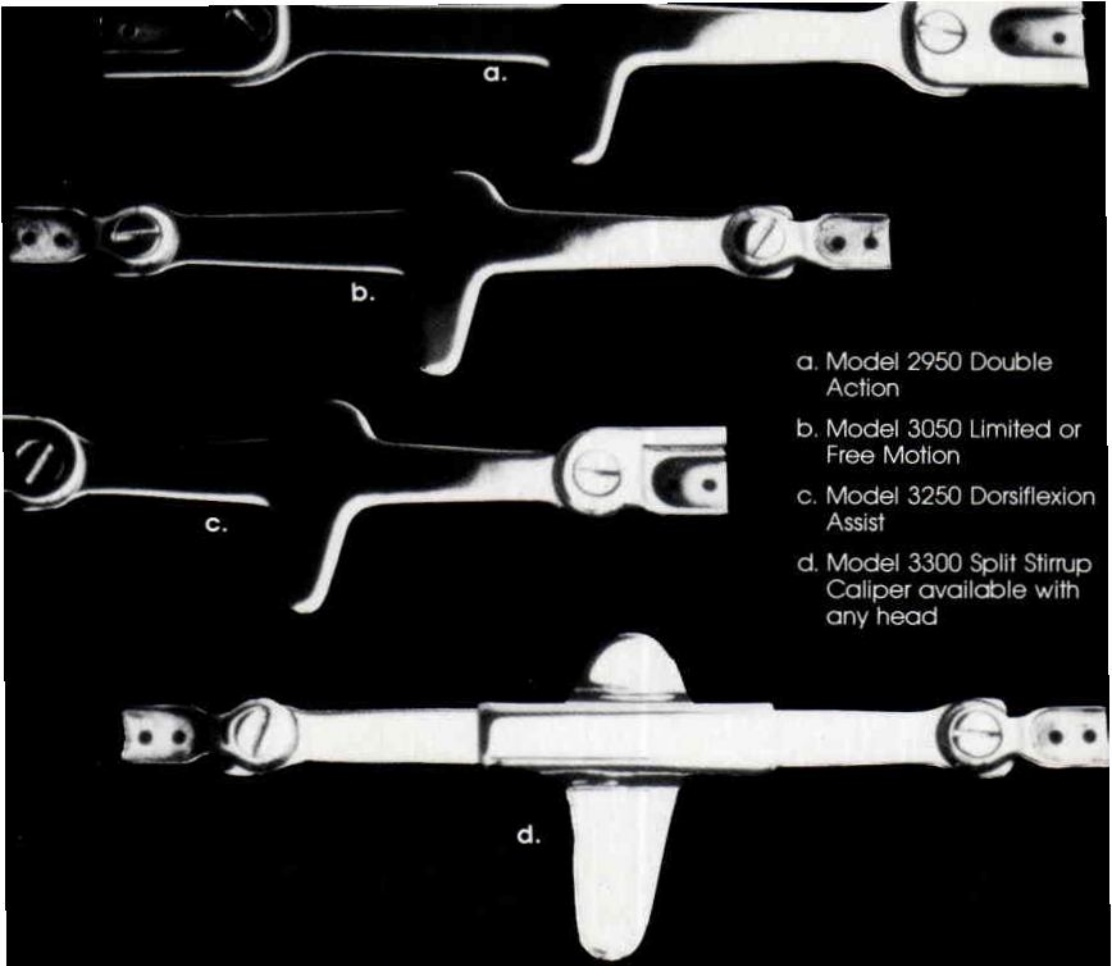

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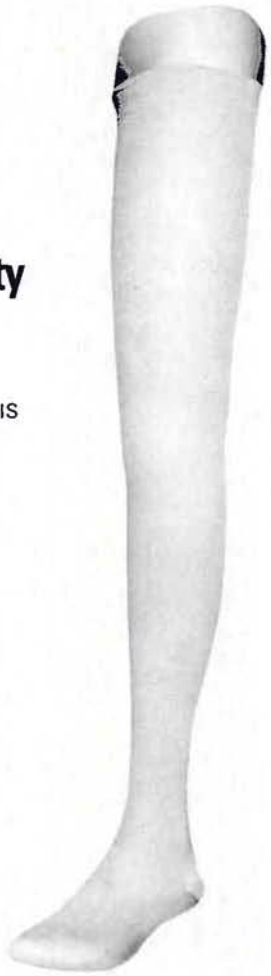

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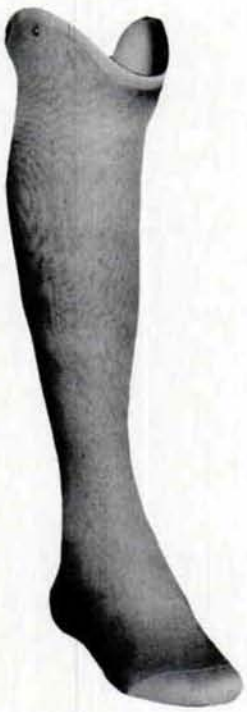

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Orthotics and Prosthetics, Vol. 35, No. 2, pp. 1-9, June 1981

Immediate Postsurgical Management of Upper-Extremity Amputation: Conventional, Electric and Myoelectric Prosthesis[1]

J.H. Malone, M.D., F.A.C.S.[2]
S.J. Childers, B.A.[2]
J. Underwood, O.T.[3]
J.H. Leal, C.P.[2]

Approximately 6-10,000 upper-extremity amputations are performed each year in the United States (6,9,18). Successful rehabilitation after upper-extremity amputation has significant economic impact both to the amputee and society, since most upper-extremity injuries occur in young working males (5,11).

Although the success rates for fingertip, finger, and hand replantation exceed 90 percent in most centers, the success rate for replantation above the wrist averages 50 percent or less in most institutions (2,4,8,13,14).

The current success rate for rehabilitation after upper-extremity amputation with fitting of conventional prosthetic devices is 50 percent or less (1,3,5,7-9,15). Although slightly more than 100 cases of rapid or immediate postsurgical prosthetic fitting after upper-extremity amputation have been reported, there appears to be a relative lack of interest in this area (9). In addition, although there have been many significant improvements in the quality, function, and reliability of externally powered upper-extremity prosthetic devices during the last five years, most of this technological advancement has been lost on contemporary surgical and prosthetic care. The purpose of this report is to review our experience with upper-extremity immediate postsurgical fitting utilizing conventional, electric, and myoelectric components.

MATERIALS AND METHODS

PATIENT DATA

Between April 1, 1979 and December 31, 1980, twelve patients with traumatic or elected upper-extremity amputations were treated with immediate postoperative prosthetic components (Table 1). The etiology for upper-extremity amputation included trauma (eight), stroke (two), brachial plexus injury (one), and tumor (one). The one patient with a brachial plexus injury required simultaneous shoulder arthrodesis in order to obtain shoulder stability for elbow control.

During this same period seven additional patients, all of whom were prior users of conventional prosthetic devices, were fitted with electric and myoelectric components for testing and evaluation (Table 2).

TABLE 1
TRAUMATIC AND ELECTIVE UPPER-EXTREMITY AMPUTATION

Amputation Level	#	PROSTHETIC TREATMENT: Immediate Postsurgical Prosthesis			
		Conventional	Electric	Electric Myoelectric	Myoelectric
Below Elbow	4	—	2	2	—
Above Elbow	7	3	—	2	2
Forequarter	1	—	1	—	—
	12	3	3	4	2

TABLE 2
CONVERSION FROM CONVENTIONAL
TO EXTERNALLY POWERED PROSTHETIC COMPONENTS

Amputation Level and (Number)	Employed	Use Prosthesis Work/Recreation*	Prefer Power**
Wrist Disarticulation (1)	1	1/1	1
Below Elbow (3)	3	3/3	3
Above Elbow (3)	3	3/3	3
			7/7 100%

*Use conventional for water and rugged activities
**See text for comments on dependability

SURGICAL TECHNIQUE

Although maximum limb length was usually preserved, amputation stumps were modified as needed *at the time of surgery* to permit optimal fitting and fabrication of the prosthetic components to be used. All amputations were closed primarily when possible, and a myoplasty, myodesis, or both were performed in all cases. The postoperative prosthetic devices used immediately after surgery were constructed so that they could be removed to allow frequent wound inspection and permit rapid prosthetic modification or repair (Fig. 1).

PROSTHETIC TECHNIQUE

Standard immediate postoperative prosthetic techniques (IPOP), as utilized in lower-extremity amputation, form the basis for the initial upper extremity cast (11,12). The IPOP cast consisted of Owen's silk against the skin as a separating agent, lamb's wool for distal stump padding, felt pads for relief for bony prominences, a Spandex stump sock, and Elastiplast for construction of the outer plastic shell. An attempt was made to provide cosmesis in constructing the original plaster prosthesis at all amputation levels (Fig. 1). Below-elbow prosthetic devices were attached to the IPOP cast using a flexible acrylic-laminate shell which was secured to the cast with tape or plaster or both (Figs. 2 and 3). Above-elbow prosthetic devices were attached to the IPOP cast through the use of a flexible polypropylene sleeve which was secured to the elbow turntable by metal band clamps and the IPOP cast with tape and/or

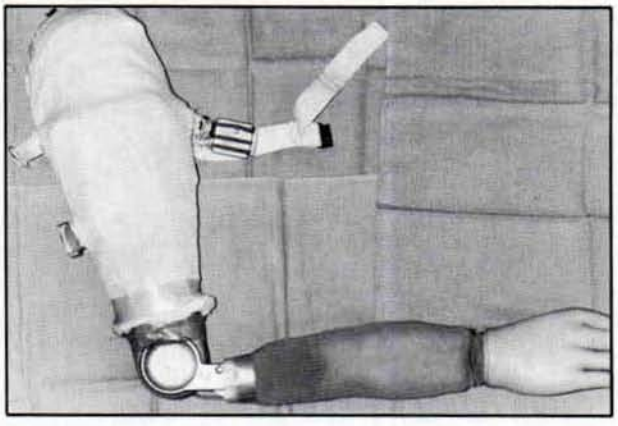

Fig. 1. Standard above-elbow Liberty Mutual/Otto Bock removable prosthesis.

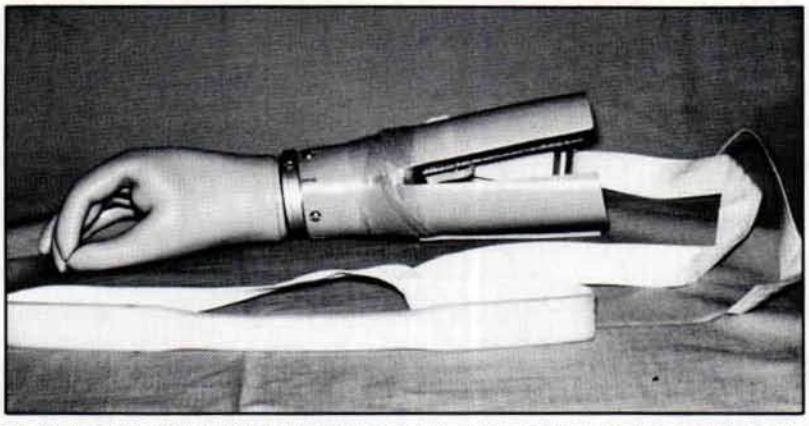

Fig. 2. A preassembled below elbow immediately postoperative prosthesis with electric hand and flexible acrylic laminate forearm.

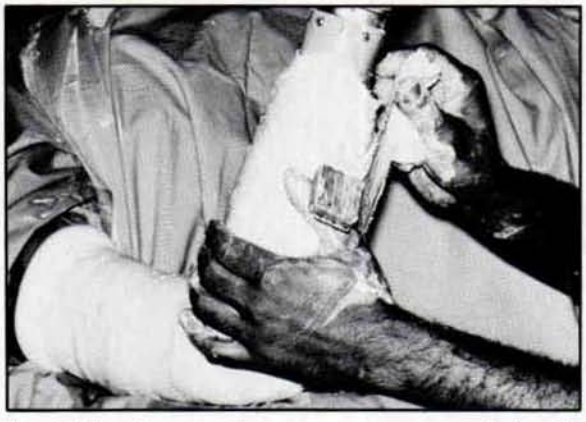

Fig. 3. Below-elbow prosthetic components are attached to the IPOP cast via flexible laminate acrylic shell with either tape or plaster.

plaster (Figs. 4, 5, and 6). All IPOP prosthetic limbs were constructed to allow interchange of hand and hook.

Our initial patients were fitted with components with either electric or myoelectric controls, but we now fit our patients with components that have been adapted for *both switch* (electric) and *myoelectric* control systems.

Below-elbow IPOP casts are locked at 90 degrees of flexion to obtain a self-suspending below-elbow prosthesis. A single axillary harness is used for switch and/or cable control.

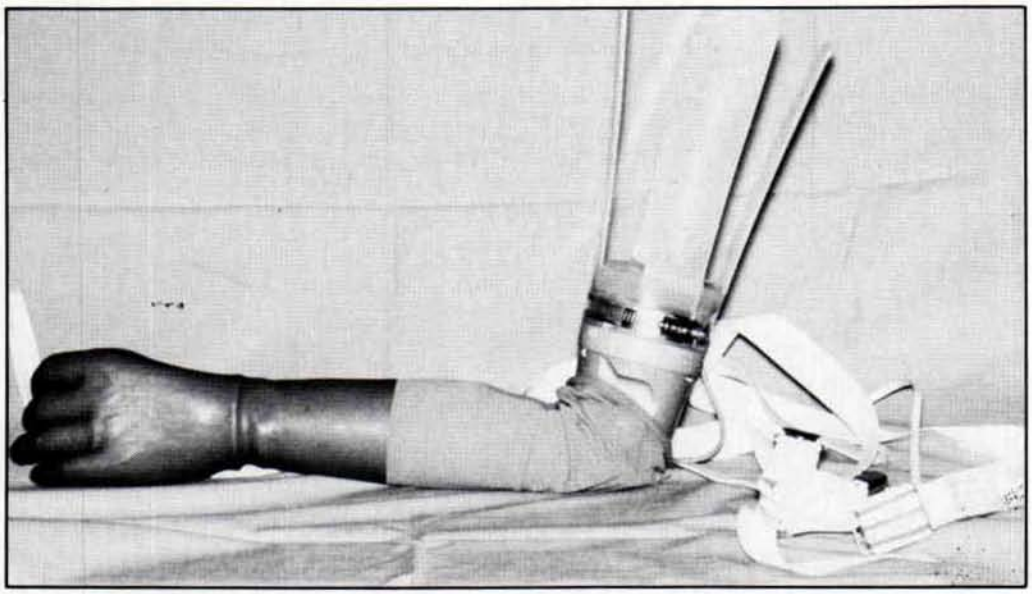

Fig. 4. A preassembled above-elbow immediate postoperative prosthesis.

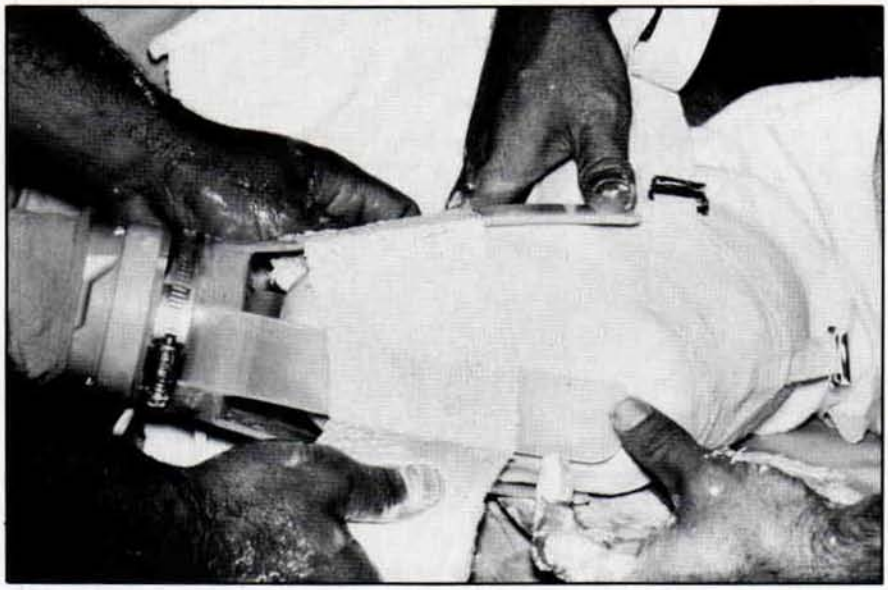

Fig. 5. The above-elbow prosthesis is attached to the prosthetic cast by incorporating the flexible polypropylene sleeve into the IPOP cast after appropriate length and position adjustment utilizing tape and/or plaster.

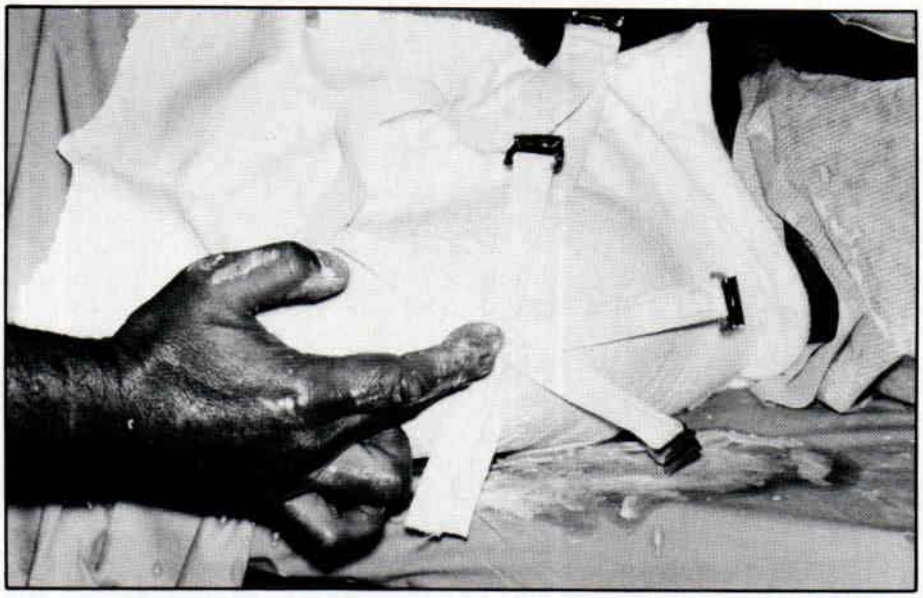

Fig. 6. In order to maintain adequate above-elbow suspension, the location of our three point harness is crucial and the usual location of the harness center point is shown in this photograph.

Harnessing for above-elbow amputees usually involves a figure-of-eight harness with axillary strap, but is individualized as required. Switch control is obtained from any point on the harness which will allow maximum patient function.

Myoelectric control points for all amputation levels are determined by standard myotesting procedures with the Otto Bock myotest unit.

For all amputation levels, switch control mounting sites are located carefully for hand and elbow components so that the standard physical motions required for utilization of a conventional hook and cable prosthesis are the same as required to utilize our electric components. When this type of standardized approach to training is used, all patients fitted with IPOP electric or myoelectric components can also be fitted with conventional hook-and-cable components and thus achieve immediate function of the conventional devices while waiting for their first temporary powered prosthesis which is provided in 4-6 weeks.

In general, we do not fit myoelectric prosthetic devices with electrodes at the time of surgery, but wait until the IPOP cast has dried for 24 hours or the time of first cast change (7-14 days postamputation). This approach has been adopted in order to avoid water damage to the electronic components during casting and to allow time for myotesting and determination of adequate electrode sites.

Our basic prosthetic components include the Liberty Mutual "Boston" myoelectric elbow which has been modified to allow switch control, United States Manufacturing/Fidelity Electronics/VANU electric 12-volt hand and the Otto Bock electric/myoelectric 6-volt hand.

OCCUPATIONAL THERAPY

Occupational therapy is begun on the first postoperative day and is continued on a daily basis throughout the patients' entire hospital course. The primary goal of occupational therapy is accelerated training of the patient to use the prosthesis, which is followed by training in the use of the prosthetic device in a two-handed manner for activities of daily living and job training, rather than teaching one-handed skills and use of the prosthesis as an assistive device.

CENTER APPROACH

All patients are treated in a dedicated amputation rehabilitation center (9-12). While in the hospital, all patients are seen daily by all members of the amputation-rehabilitation team. When discharged from the hospital, pa-

tients are followed longitudinally by all members of the team. Outpatient care is organized by and is under the direction of the amputation-rehabilitation team coordinator (SJC). Our program coordinator also supervises the educational training that the patient and his family receive during the pre- and postoperative periods as well as provide liaison between our program and other community agencies.

RESULTS

All wounds that were closed primarily healed primarily (12/12, 100). None of the patients treated with immediate postoperative prosthetic casting techniques had any injury to the amputation stump due to the cast or the immediate fitting of prosthetic devices.

There were no operative deaths and no acute morbidity. One late stump revision (1 yr) was performed for ectopic bone formation.

All three above-elbow amputees fitted with immediate postsurgical conventional hook-and-cable prosthetic devices had complete function with the ability to lock the elbow in all positions as well as open and close the terminal device in all positions within an average of seven days (range 2-14 days). Patients fitted with immediate postsurgical electric hands and elbows were functional after 10-15 minutes of training, and patients fitted with myoelectric elbows and hands were functional within 12-48 hours after initial training and practice.

At one month after amputation, all of our patients were wearing their new prosthetic devices effectively (12/12, 100%). Longterm followup shows the following: one above-elbow amputee died six months after hospital discharge in an automobile accident; however, at the time of death he was wearing his prosthesis 12-14 hours a day and using it for work and activities of daily living; one patient has quit wearing his forequarter prosthesis because of complicated medical and psychological problems which have required repeated hospitalization; all other patients (10/10, 100%) continue to wear their prosthetic devices 8-18 hours per day (average 12 hrs) and use them in all activities of daily life and work.

The rate of employment after traumatic or elective upper-extremity amputation for our group of patients is demonstrated in Table 3: Four of six patients employed before amputation have been successfully reinstated at their same job after amputation; two patients are in training for new jobs; three patients were students when injured and all three have returned to school; two patients were in retirement at the time of injury and have gone back to their postretirement activities; and one patient who was unemployed when injured was temporarily rehabilitated and employed. All employed patients use their externally powered prosthetic devices rather than their conventional prosthetic components for work and most activities of daily living. Time from amputation to work ranges from one week to one year (mean 3.8 months), and the length of time at work ranges from 3-18 months (mean 8.4 months) (Table 3).

TABLE 3
TRAUMATIC AND ELECTIVE UPPER EXTREMITY AMPUTATION
RATE OF EMPLOYMENT

Amputation Level and (Number)	Employment Before Amputation	Employment After Amputation	Job Training	Student	Retired	Time From Amputation To Work (Months)	Length of Time at Work (Months)
Below Elbow (4)	3	1	2	1	—	1	18
Above Elbow (7)	3	3	—	2	2	3,12,m	3*,6,12
Forequarter (1)	—	1	—	—	—	3	3**
	6	5	2	3	2	Mean = 3.8mo	Mean = 8.4mo

*Death 6 months after amputation
**Quit wearing at 6 months, unemployed prior to amputation

None of our surgical patients have developed painful phantom syndromes. In fact, many of these patients have transformed all sensory feeling from their phantom limb to their prosthetic device. This *sensory transformation* of phantom sensation to the prosthetic limb applies only to those patients fitted with immediate postsurgical prosthetic devices (fitted within *one month* of amputation) and was not seen in our seven patients who were fitted months to years after their original amputation.

Evaluation of our seven amputees who were prior wearers of conventional prosthetic devices and then provided externally powered prosthetic components is shown in Table 2. All seven patients are employed and all preferentially use their externally powered prosthetic devices for work, recreation, and/or activities of daily living.

Comparison of hand-versus-hook function by patients suggests that electric and myoelectric prosthetic hands provide increased function when compared to standard prosthetic hooks. Those patients doing heavy manual labor had some difficulty with their electric hands because of component failure and breakage. A few of these patients have returned to using their hook for heavy work; however, none of these patients want to give up their hand for light work, social functions, or activities of daily living.

As might be expected, all patients are extremely pleased with the cosmetic value of their externally powered components.

DISCUSSION

The success rates for rehabilitation after upper-extremity amputation vary with the quality of surgery, type of prosthetic fit, quality of fit, and the patient. In general, the success rates for rehabilition after upper-extremity amputation are highest when the patient is fitted as rapidly as possible after injury (3,5,7,16-20). Conventional (soft or rigid dressing without fitting the terminal device until after complete wound healing and complete stump maturation) upper extremity-amputation and rehabilitation often result in late fitting of upper-extremity amputees. By the time an amputee is fitted with a prosthetic device in most settings where mean delivery time of the prosthesis is six months (6), he has become skilled at being a one-handed individual and sees very little use for "assistive" prosthetic devices.

In a previous publication we noted that the overall rehabilitation data on 109 published cases of rapid and/or immediate postsurgical fitting for upper-extremity amputation documented a rehabilitation time which averaged ten days, a mean fitting time for permanent prostheses of approximately 12 weeks, and most importantly, an overall amputee rehabilitation rate greater than 90% (100/109) (9). During the past 20 months, our group has fitted all levels of traumatic upper-extremity amputation from below-elbow to forequarter with immediate postoperative conventional, electric, and myoelectric prosthetic devices. In addition, we have performed several elective above-elbow amputations with the fitting of externally powered devices in patients with neurologic dysfunction of their arm due to stroke or brachial plexus injury. All of our patients had rapid prosthetic function with rehabilitation times ranging from 10-15 minutes to 2 weeks, depending upon the type of device which was fitted to the patient. In general, the rehabilitation times for patients fitted with externally powered upper-extremity components is significantly less than that required for patients fitted with conventional hook-and-cable prosthetic devices.

Although our patient group is small and our data are preliminary and further longitudinal evaluation is required, our data suggest that patients fitted with immediate postoperative prosthetic devices and who were employed prior to amputation, can continue to be employed after amputation at their prior job in most cases. Several of our patients have had to undergo job training and have not been able to return to their original job, not because of physical incapacitation, but because of either specific instruction from legal counselors or because of local union-employer agreements regarding disability and rehabilitation.

The patients we have studied who have been conventional prosthetic users prior to receiving externally powered prosthetic components all prefer externally powered prosthetic devices for all but heavy work or water related recreational activities. In addition, all of our patients fitted with immediate postoperative prosthetic

components and then supplied with both conventional and externally powered prosthetic devices after rehabilitation prefer their externally powered components for most activities.

All of our patients who were provided with both an externally powered hand and a cable controlled hook prefer the electric/myoelectric hands for almost all activities.

We believe that, if possible, the amputation stump should be closed primarily with a myoplasty and/or myodesis technique. Adequate muscle stabilization is especially important when a myoelectric prosthetic device is to be fitted to the patient. In addition, we believe the amputation stump should be modified as required at the time of initial surgery in order to allow the best possible fit of the planned permanent prosthetic devices.

Early in our series, three above-elbow amputees were fitted with conventional prosthetic components. We no longer utilize conventional prosthetic components for immediate postsurgical fitting due to prolonged rehabilitation time with respect to prosthetic function compared to externally powered components and because our patients achieve greater functional capability through the externally powered components than is possible through the use of conventional hook-and-cable prosthetic devices.

When comparing stump wrapping after amputation of the upper-extremity and late fitting of conventional prosthetic devices to immediate postoperative fitting we believe that there are multiple advantages to immediate postoperative prosthetic application of either conventional or externally powered prosthetic devices for upper-extremity amputation including control of edema, decrease in postoperative pain and phantom limb sensations, accelerated wound healing, improved stump stability, enhancement of stump maturation, rapid patient rehabilitation, decreased length of hospital stay, decreased hospital costs, improved psychological outlook of the patient, increased prosthetic use, and maintenance of some type of proprioceptive input through the amputation stump.

In using externally powered upper-extremity devices it is important that the total weight of the final permanent prosthetic arm does exceed the weight of the amputated limb. In our experience it is possible to construct both myoelectric below-elbow prostheses and electric/myoelectric above-elbow prostheses (with power hand and 2 battery packs) which weigh the same or less than the amputated limb (2-2lbs, below-elbow, 5-5lbs, above-elbow). In addition, ancillary suspension techniques such as suction socket, partial suction socket, and Silastic sleeve suspension are all great aids in achieving increased patient range of motion and decreased overall prosthetic weight.

Most externally powered prosthetic devices have been durable and reliable. We have had few problems with the Liberty Mutual "Boston" myoelectric elbow, the Otto Bock electric/myoelectric 6-volt hand, and the United States Manufacturing/Fidelity/VANU 12-volt hand; however, the US Manufacturing/Fidelity Electronic/VANU electric elbow and electric elbow-hand combinations have been very disappointing and we no longer use those externally powered devices.

It is our overall feeling that further investigation into rapid rehabilitation and improved patient function with immediate postsurgical application of externally powered prosthetic devices for upper-extremity amputation is definitely warranted due to the potential advantages to the patient and society.

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Footnotes

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Prosthesis Management of the Cancer Patient With High Level Amputation[1]

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Prosthetics and orthotics management of patients who have suffered from major surgery due to cancer has, in the past, not addressed the unique problems associated with this disease. For these patients, one must consider not only the physical effects of amputation or other major surgery, but the psychosocial situation as well. Amputations due to cancer usually differ from other kinds of amputation in the severity, i.e., massiveness, of the amputation. This and the insidiousness of the disease obviously affect patient management.

This report represents part of the results of a comprehensive effort to establish an aggressive patient management program which deals with the prosthetic/orthotic needs of the patient. Psychosocial aspects were studied also.

Objectives

The objectives of this research were the design, development, testing, and utilization of a cost-effective limb prosthesis/orthosis system for people who have undergone amputations/tissue resections due to cancer. The specific amputation levels considered for which respective devices were designed included hip disarticulation, hemipelvectomy, and forequarter. Another objective was the design of orthoses for lower limb tissue resections.

The total rehabilitation process for this type of cancer patient was to include an aggressive early intervention program involving several pertinent professional disciplines. This was believed to help dissipate the psychosocial consequences of radical surgery.

Special Considerations for the Cancer Patient

Cancer is a disease that is associated with uncertainty and fear on the part of the patient. Unlike many other diseases that are specific to certain portions of the body, this one is insidious and frequently defies containment. The individual, on being labeled with the term "cancer," is frequently demoralized and confused as regards, to say the least, his future employment. This person must be made to know from the very onset that many disciplines are working together and are committed to his resumption of a lifestyle that is as normal as is practicable.

The amputation levels selected for the prosthetics research (Chart 1) were chosen in part

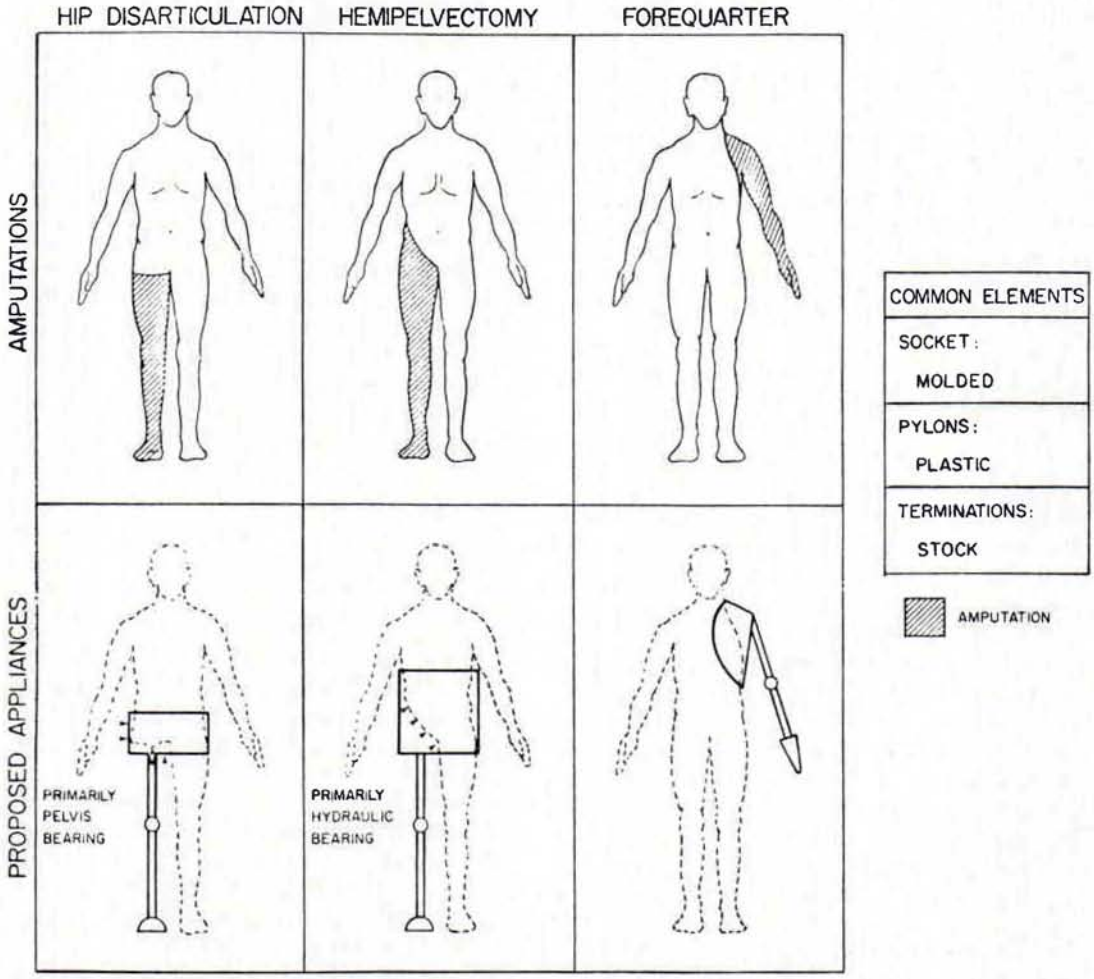
HIP DISARTICULATION
HEMIPELVECTOMY
FOREQUARTER
AMPUTATIONS
PROPOSED APPLIANCES
PRIMARILY PELVIS BEARING
PRIMARILY HYDRAULIC BEARING
COMMON ELEMENTS
SOCKET: MOLDED
PYLONS: PLASTIC
TERMINATIONS: STOCK
AMPUTATION

CHART I

because of their prevalence in cancer care and because when amputations are at such a high level, the probability of patient acceptance of a device is decreased. This is a result of the massiveness of the appliances, discomfort, and resultant difficulties of control. Control can be enchanced by reduction of weight, and, in the case of forequarter amputation, by myoelectric control.

Early fitting through direct molding of a thermoplastic prosthetic socket for lower-limb amputees does, in part, mitigate the discomfort factor by reducing edema. It allows quick fitting and early mobilization of the patient with a preparatory prosthesis.

Plastic pylons are lighter, less expensive, and more readily adjusted than a comparable unit of aluminum which is presently used in modular prostheses.

In spite of early prosthetics management, high level lower-limb amputees may not wear their prostheses because of discomfort caused by the socket and the difficulty in controlling a biarticular system which hip-disarticulation and hemipelvectomy prostheses constitute. Thus, design and development of an improved socket and control system was another important consideration.

The application of modularity in orthotics for cancer patients is expected to result not only in a cost reduction if these components can be mass-produced, but more importantly, it permits immediate fitting of orthoses. This allows rapid diagnostic evaluation of the type of orthosis required and results in an improved delivery system.

This report includes only our work with the hip-disarticulation and hemipelvectomy cases.

Prostheses for Hip-Disarticulation and Hemipelvectomy Cases

PATIENT REFERRAL AND ORIENTATION

The first nine months of actual patient participation, beginning in May, 1978, involved referrals from within N.Y.U. Medical Center by the patient's physiatrist. Due in part to the death of Dr. K. Francis, the original consultant orthopedic surgeon on the project, patient referral diminished. It was for this reason that an official agreement between N.Y.U. Medical Center and Sloan-Kettering Memorial Hospital was established in April 1979.

Most of the participants were in-patients at the time of evaluation for candidacy and initial fitting.

Patients were referred to the project either by the surgeon or through the recommendation of the physical therapist from Memorial Hospital, with the approval of the surgeon and/or physiatrist.

Except for a few out-patients, most patients (plus family members for pediatric patients) were seen bedside at Memorial Hospital for orientation. The project was explained to the patient, stressing the value of early ambulation with bilateral weight-bearing. The fitting procedures using direct molding of a thermoplastic material and the general time-sequence and approximate number of sessions needed for fabrication and fitting were discussed. In addition, the experimental nature of the project was stressed. Patients were told that problems with technique and materials could arise. To augment the orientation session, slides depicting the exact molding technique were shown with detailed description. The intention was to eliminate any apprehension and confusion prior to the actual fitting sessions.

Considerable time was spent with patients in the orientation session, often repeating and reinforcing the important facts about the project. During their actual participation in the project, however, several patients still had misconceptions and questions which had previously been discussed. To augment the conveyance of the initial information and ensure its reinforcement, a patient brochure was written. The questions and information discussed in this brochure reflect some of the common misunderstandings and points that the patients did not grasp.

When the patient agreed to participate in the project, he was asked to sign consent forms, one for Memorial Hospital records, and one for NYU Medical Center records. At that time, basic measurements were taken of the amputation site. These were used as guidelines for cutting a blank thermoplastic pattern for the socket of the preparatory prosthesis to be fitted at a subsequent session.

In addition, pertinent personal data were noted. The medical chart was also consulted for diagnosis, past and present medical history, and possible regimen of chemotherapy or radiation, if already planned, as this would affect the patient's schedule and tolerance, as well as the type of preparatory prosthesis to be fitted.

Depending on the patient's physical stamina and endurance, the protocol for one or more sessions was decided upon. If the patient did not demonstrate any physical limitations hindering his/her endurance, fabrication and fitting the prosthesis all in one session was to be attempted.

PREPARATORY PROSTHESES

To accomplish one of the major objectives of this research, i.e., early ambulation through cost-effective means, this particular aspect of the project was concerned with the search for a material which would allow direct molding of a socket on the patient. The conventional method is not only time-consuming but costly, since the conventional socket is plastic laminated over a plaster mold from an impression of the patient's body. Again, to minimize cost and to permit ready adjustability, a pylon-type prosthesis utilizing a thermoplastic tubing was designed. An alternative to a direct molding socket was the design of a ready-made totally adjustable socket.

Molding the Socket

Aquaplast™[3] was found to be the most suitable material for the purpose described. While a number of other materials have been experimented with earlier in the project, they are not described in detail here since the results of such experimentation have been included in progress reports.

Aquaplast is a low-temperature thermoplastic polyester which softens in 60 deg. C (140°F) tap water. Moldability is evident as the material turns from a cloudy opaque appearance to transparency. Its elastic properties allow stretching in any direction. It is also self-bonding. Inconsistencies in the molding properties of Aquaplast due to the amount of stretch and stickiness caused some initial molding problems. After appropriate follow-up with the manufacturer, however, most of the problems were alleviated.

From the measurements taken during patient orientation, a socket pattern is cut from the Aquaplast sheet material. The molding procedure takes place with the patient standing between parallel bars. Cotton stockinette is pulled up and over the patient's amputation site and body to protect the skin from the stickiness as well as the heat of the material. Pressure relief felt pads are placed over the iliac crest(s) extending to the anterior superior spine on the sound side and involved side, using medical adhesive spray.

A hydrocollator is used to heat the Aquaplast until clear. Once it is softened and completely transparent, it is draped and stretched over the body. The ends can be closed together by pinching anteriorly, as the material sticks to itself. Precise molding must be accomplished at this point in a relatively short time because the Aquaplast cools and sets quickly. Thus, an additional person may be needed. Using an elastic webbing around the waist just superior to the iliac crests, pulling forward and downward, proved to be the most effective way to ensure accurate contouring (Fig. 1). Molding the distal

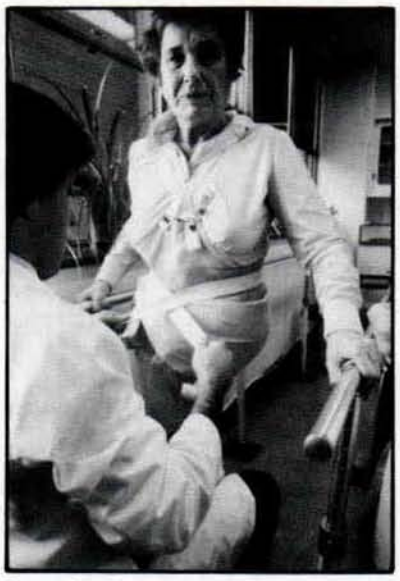

Fig. 1. Preliminary steps in molding for the hip-disarticulation socket.

end of the socket is done effectively by pulling a length of 12.5 cm or 15 cm (5 or 6 in.) wide piece of cotton stockinet diagonally toward the iliac crest of the sound side (Fig. 2). The

Fig. 2. Molding for the hip-disarticulation socket.

Fig. 3. Marking reference lines and trim lines on the Aquaplast cast.

stockinet must be sufficiently long to tie a knot just superior to the anterior superior spine. This establishes and imparts a crucial anatomical dimension in the socket. It optimizes the interface and thus control of the prosthesis. Care must be taken to eliminate and/or smooth out all irregularities, as the Aquaplast tends to shrink as it cools. During this time, the patient is instructed to stand tall and erect in a comfortable position, with both iliac crests level.

When the Aquaplast cools and becomes opaque again, trim lines and reference lines for prosthetic alignment are marked. (Fig. 3) In the frontal plane, a plumb line is marked on the socket at a position under which the prosthetic foot will be placed. This is usually a point 7 to 8 cm from the center of the heel of the sound foot. In the sagittal plane, a plumb line is projected on the socket in line with the patient's medial malleolus.

The socket is then removed from the patient by cutting it anteriorly with heavy-duty surgical scissors or a cast cutter, if necessary.

The procedures described above are practically identical for hip disarticulations and hemipelvectomies, with the exception that with a hemipelvectomy patient, the superior trim line of socket is somewhat higher than for a hip disarticulation, extending to the level of the xiphoid process. Even though the crest of the ilium may be missing in the hemipelvectomy patient, the elastic webbing is still used in the same manner to impart a waist level on the amputated side for cosmetic reasons and to enhance suspension.

Pylon

Polyvinylchloride (PVC) tubing is used for the pylon and is attached distally to a solid-ankle cushion-heel (SACH) foot, and proximally at the measured height to the Aquaplast socket (Fig 4). For lightweight and pediatric patients, 1-inch (I.D.) PVC is used. For heavier patients, 1¼-inch (I.D.) tubing is needed. PVC tubing is available commercially from plumbing supply houses since it is widely used as cold water piping. The proximal end of the PVC tubing is split in such a way as to produce three wings of approximately 15-20 cm length. Heating these wings allows easy contouring to the Aquaplast socket at a point where the center of the tubing coincides with the plumb lines previously marked on the socket. The wings, and thus the pylon tubing, can now be attached securely to the Aquaplast socket by means of a heated Aquaplast patch of appropriate size to cover and extend beyond the wings of the tubing as well as the proximal 5-7 cm of a specially-made PVC plug which fits into the tubing and is held in place with a hose clamp.

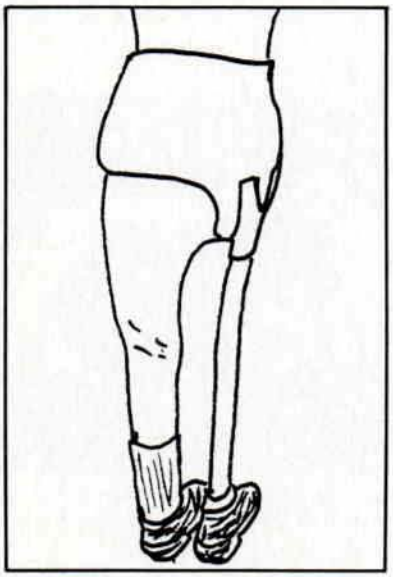

Fig. 4. Schematic of first stage prosthesis. No articulated joints are used.

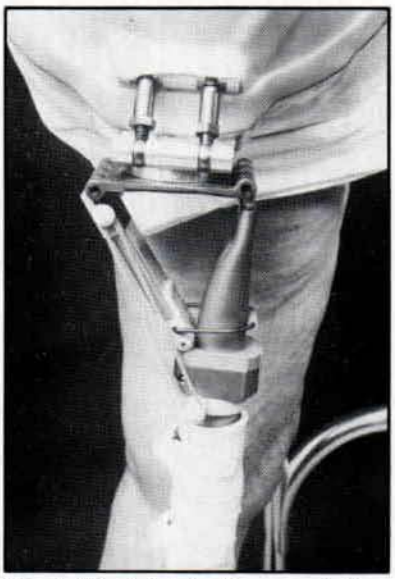

Fig. 5. Hip joint in the second stage prosthesis.

Any rough edges on the trim line of the socket can be smoothed with sandpaper. Velco closures are attached across the anterior opening in a criss-cross fashion to minimize socket displacement on weight-bearing and in the swing phase of walking.

It should be noted that this type of pylon does not include a hip joint or knee joint. The reasons for this design are discussed under Patient Application.

Pylon with Hip Joint

A pylon incorporating an Otto Bock No. 06-7E4 model hip joint is used in the second stage of the prosthetics management of high level lower-limb amputees. To permit the patient to sit with greater comfort and ease and to allow an improved, relatively safe, gait pattern. It is applied approximately two weeks after the fitting of the first pylon, on the assumption that by that time sufficient stump changes have taken place to warrant socket replacement and the patient will have mastered the use of the first pylon.

Fit and alignment are identical to the first pylon fitted. The placement of the hip joint is the same as that for Canadian hip-disarticulation prostheses. Distally the hip joint is adapted to the PVC tubing, and proximally it is attached to the socket by means of a specially designed disc with adjustment capability for abduction and adduction (Fig. 5). Attachment to the socket itself is by means of monel straps held to the socket by a patch of Aquaplast as described in the previous section. Change in alignment, other than abduction or adduction, can be accomplished through the excellent thermoplastic characteristics of PVC, i.e., heating the tubing for making the desired change in alignment.

Prefabricated Adjustable Socket with Pylon

Early in the project it was believed that many patients would be undergoing chemotherapy and radiation therapy and, thus, would present severe volume changes in the amputation site and delayed wound healing. For that purpose, a prefabricated adjustable socket and pylon was designed. This preparatory prosthesis consists of a prefabricated plastic molded pelvic girdle, commercially available for Milwaukee braces and low-profile Boston spinal orthoses. The girdle is adapted to PVC tubing and SACH foot (Fig. 6). To provide weight-bearing, an adjustable hammock is inserted in the girdle under the amputation site (Fig. 7). The hammock not only provides adjustment for volume changes but consists of a throw-away material, e.g., sterilized stockinette or towel. Circumferential adjustment is possible through the posterior

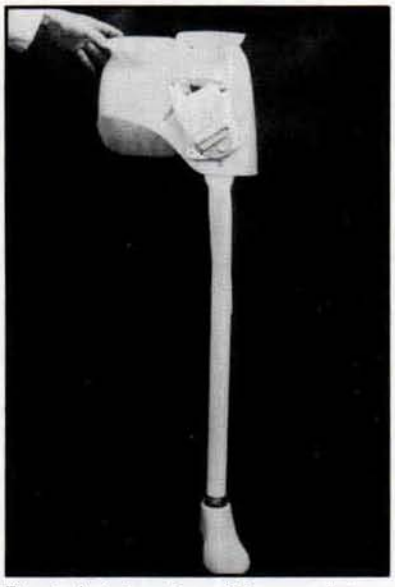

Fig. 6. Anterior view of the second stage prosthesis with prefabricated adjustable socket.

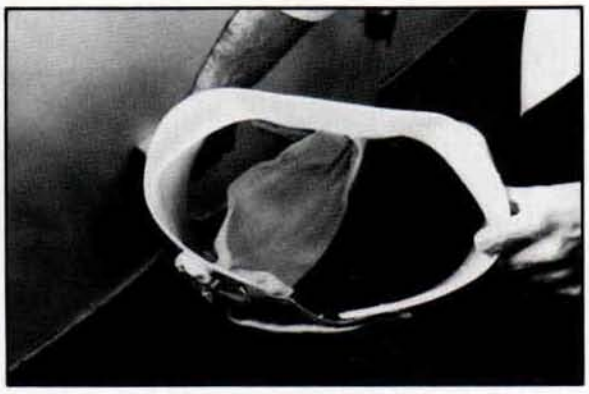

Fig. 7. Top view of prefabricated adjustable socket showing the adjustable hammock that provides weight-bearing

opening in the girdle. This design was tested on a hip disarticulation amputee who was, indeed, able to ambulate on this preparatory prosthesis. As the project progressed, however, it was found that these patients could be managed just as well with the Aquaplast socket.

Patient Application

Protocol of Prosthetics Management

As stated, one of the fundamental goals of the project was to fit cancer amputees with a temporary device as soon after surgery as medically feasible. The protocol of early prosthetic intervention evolved as follows. Once the patient is orientated and medically cleared to participate, the fabrication and fitting procedures are carried out in the manner described above. The first socket is attached to a pylon and SACH foot, without a hip or knee joint. Both joints are excluded from the design of the first socket to insure optimal stability for the patient and expedience in fabrication. The entire procedure, from molding of the socket to attaching the pylon and foot, to final alignment can be accomplished in three to four hours. Following this, gait training is initiated. The patient is asked to stand with the prosthesis and practice equal, bilateral weight-bearing. Basic body mechanics necessary for effective ambulation are taught. In addition to a posterior pelvic tilt used to initiate swing phase, the patient also practices maintaining both sides of the pelvis level. This latter movement is accomplished by contracting the contralateral abductors and is especially valuable to master early in the rehabilitation program. When such patients are not wearing a prosthesis, the pelvis or remaining tissue on the involved side is often allowed to drop. This habit frequently becomes a comfortable and natural part of the patient's body mechanics, and is difficult to reverse when prosthetics training ensues.

Progressive gait training activities are carried out in subsequent physical therapy sessions. Ambulation is limited within parallel bars until the patient acquires skills of equal weight shifting, defined and smooth pelvic tilt, and minimal trunk deviations. The patient then progresses to using a walker, forearm crutches, or canes outside the parallel bars. As with any temporary device, using at least one cane is recommended to prevent full weight-bearing on the amputation site so soon after surgery, as well as for safety reasons.

Although this first prosthesis does not have a hip joint, a modified sitting posture is still possible. The patient is able to sit leaning back on the edge of a chair with the uninvolved side on the seat, or on a stool or high chair where less hip flexion is necessary.

After a few weeks of prosthetics use, and once safe ambulation and basic skills have been achieved, it is expected that the area about the amputation site will have fluctuated sufficiently to warrant fitting of a new socket, which can be made in the manner described above.

At this stage, of gait training, the patient is likely to be ready for a pylon with a hip joint. An obvious advantage of the hip joint is less restricted movement and greater comfort in sitting. Also, the hip flexion motion available in the swing phase improves the gait pattern. Having the essential ambulation skills already mastered, the addition of a hip joint requires only minor changes in body mechanics, for safe use of the prosthesis. Improved sitting convenience and gait pattern allows the patient to wear the second preparatory prosthesis several hours at first; then, depending on tolerance, gradually progressing to a full day. Instructions are given to be mindful of sharp red areas and pressure spots, and to report any pain or pinching. Fitting adjustments can be made readily by spot heating the Aquaplast material.

The ultimate goal of wearing this progression of preparatory prostheses is to expedite the process of accepting and utilizing a permanent prosthesis, perhaps as early as two months after surgery. Not only can early containment of body tissues in a socket prevent edema and promote healing, but the patient is more likely to become a functional ambulator with a definitive prosthesis. Current practice of prosthetics management indicate this to take at least three to six months post-surgery without a temporary prosthesis. What results is a patient who has not borne weight bilaterally for several months, and who has adjusted to a fast gait pattern with crutches and one limb. Thus, the rejection rate of the permanent prosthesis for such high level amputees is rather high. The protocol developed in this research of using a sucession of a least two preparatory prostheses serves to alleviate this problem through early intervention, by maintaining bilateral body schema in ambulation.

Results of Patient Fittings

A total of ten patients were fitted with preparatory prostheses. Five patients received full hemipelvectomy amputations (Fig. 8) two received hip disarticulations (Fig. 9), and the re-

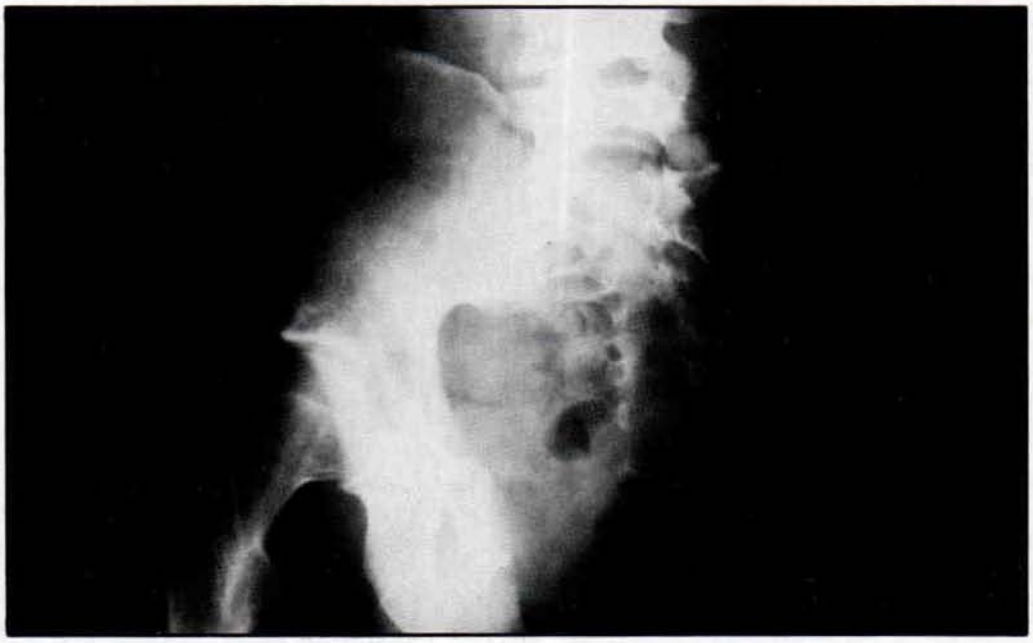

Fig. 8. X-ray picture of hemipelvectomy case

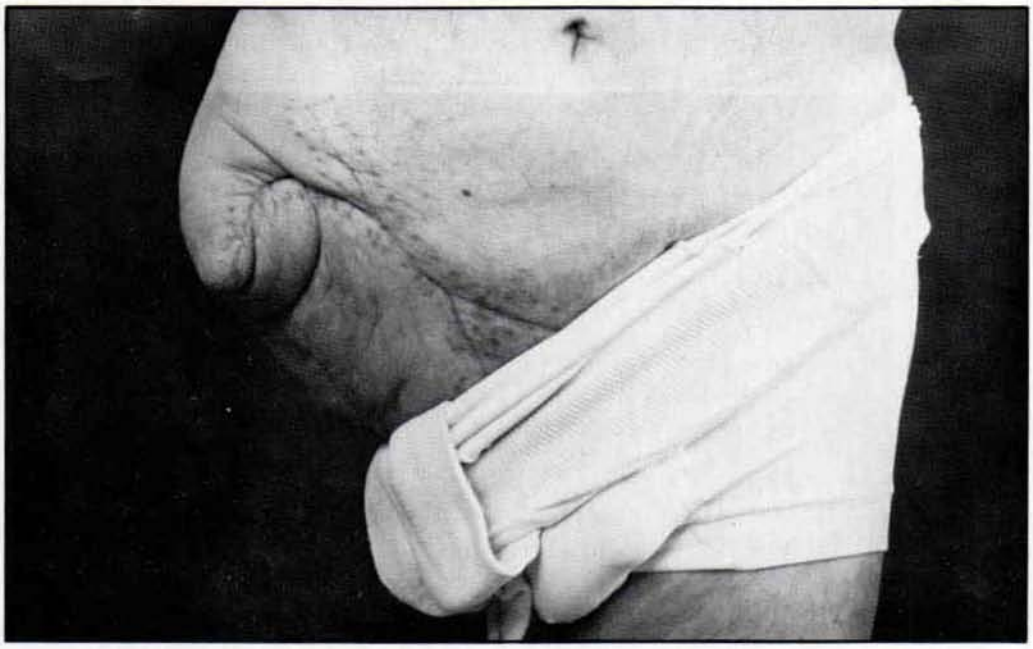

Fig. 9. Anterior view of a hip-disarticulation case

maining three received modified hemipelvectomies (Fig. 10), i.e., a portion of the ilium was preserved. Four patients were male and six, female. Ages ranged from 11 to 74 years. Four of the patients were in the 10-20-year-old range, two were in their early 30's, and the remaining four ranged from age 57 to 74.

Eight of the ten subjects received their first temporary prosthesis at an average of 16 days after surgery, some as soon as 7-9 days after the amputation. The first two patients in the project, however, were not fitted until seven weeks postsurgery, because the referral system from the surgeon at that time was still being refined. As a result, referrals for these two patients were not received until well after the surgery date. The majority of patients received two temporary sockets, the second usually having a pylon with a hip joint.

Although a preparatory prosthesis can be fitted in the hospital setting in a matter of three to four hours, in time it was realized that it is less strenuous for patient and prosthetist to change the original protocol to a two-stage fitting. Additional time was sometimes needed for realignment and for readjustment of the socket by spot heating or padding for pressure areas. When a pylon with hip joint was used, another 30 minutes were necessary. Although patients usually did not have to stand for more than 20 minutes at one time in the parallel bars, most of them nevertheless did not have the endurance or physical tolerance to stand up and sit down periodically for this length of time. Not only did most still feel generally debilitated from their recent surgery, but several of the patients were undergoing highdose chemotherapy treatments during this period. Nausea, vomiting, diarrhea and a total loss of stamina and energy are serious side-effects of such treatment. Thus, it was often very difficult to complete the procedure in one session. One of the patients did, however, bear through a four-hour session of fitting a socket and pylon with hip joint. Her satisfaction and gratification to stand and walk with a prosthesis by the end of the afternoon, make it a well-worthwhile effort, but she was clearly the exception. Most of the other patients were seen in two sessions: the first for socket fabrication, the second for final fitting and alignment. Between these sessions, the pylon with hip joint and socket were assembled in the prosthetics laboratory, so that less time would be demanded of the patient. Scheduling problems arose most often because of interference with chemotherapy treatments

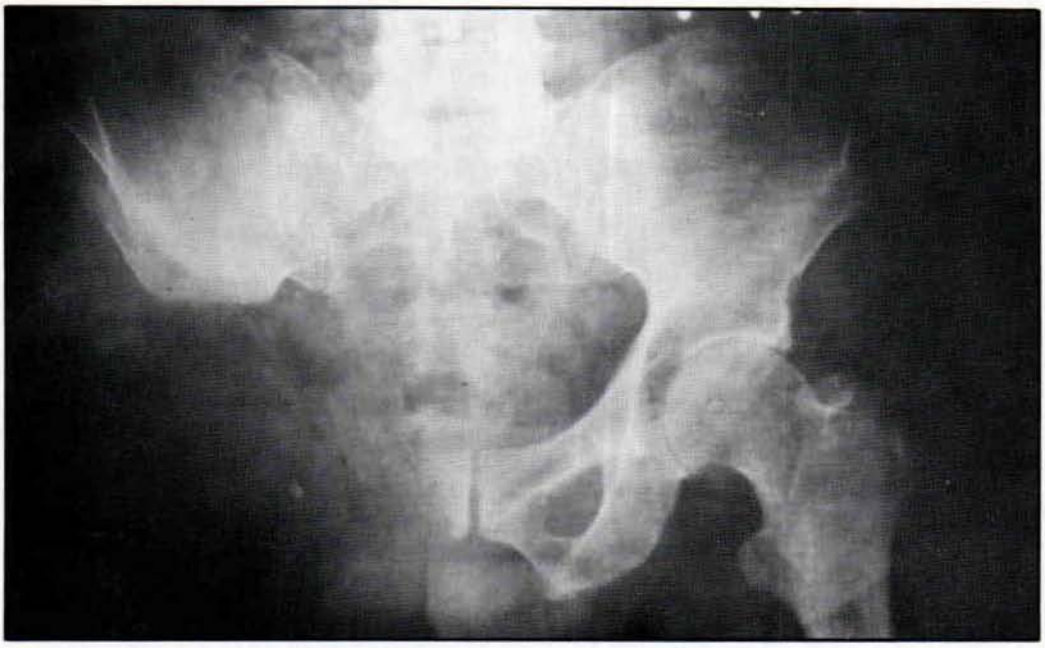

Fig. 10. X-ray of amputee in whom part of the ilium was preserved.

and the complications which resulted. Patients needed at least two days after a chemotherapy session to recover sufficiently from its effects to resume ambulation training.

Gait deviations common to the majority of patients were 1) the overflow of thoracic extension associated with the posterior pelvic tilt and 2) vaulting on the non-prosthetic side in the initiation of swing phase. Trunk rotation during prosthetic swing phase was more prevalent in patients with hemipelvectomy than in those with hip disarticulation. This is not an unexpected finding since there is a lack of bony control on the involved side; thus, these patients exhibited excessive rotation with resultant circumduction. Verbal cuing, visual feedback, and constant reinforcement reduced this deviation to an acceptable degree.

Bearing weight on the amputation site was, for the most part, painless once all uneven surfaces in the socket were smoothed out. Two patients had problems with phantom pain which interfered with training and function. One older woman reported experiencing pain after she ambulated and the prosthesis was removed. The other patient was pediatric and experienced sharp phasic phantom pains while standing in the prosthesis. Since his standing tolerance was very limited initially because of this, his training was limited for the first two weeks. Once this dissipated, training resumed at a normal rate. Only one patient had a markedly difficult time shifting weight to the prosthetic side during initial gait training sessions. She exhibited lateral trunk flexion to the prosthetic side and inadequate extension in prosthetic stance phase. This improved with more gait training. Another patient had problems with excessive medial placement of the pylon, resulting in adduction of the pylon upon weight-bearing. This was corrected by heating the PVC pylon near the socket attachment point, and simultaneously translating the tubing laterally and in abduction while the material was pliable. Once cooled, it maintained this alignment. During the course of her gait training, the same patient began to flex her sound knee excessively during the stance phase. On the basis of new measurements taken, it was evident that there was considerable shrinkage, that allowed her to slip further into the socket. This, in effect, made the prosthesis appear shorter. Hence, knee flexion compensated for the leg length discrepancy. A new socket molding was, therefore, indicated.

Early in the project, it was assumed that

shrinkage would be a common phenomenon for several weeks after surgery. This would precipitate the molding of a new socket two or three weeks after the first one, at which time a hip joint could be attached. In actuality, significant shrinkage occurred in only two patients, both female, in their seventies, and overweight.

The five youngest patients, ages 11 to 32, gradually gained weight during the weeks following surgery, despite short intervals of rapid weight loss immediately following chemotherapy treatments. Thus, rather than shrinkage, it was actual weight gain in half the patients that necessitated change of socket within three weeks postsurgery.

Only one patient did not receive a hip joint in the "second phase" prosthesis. Her prognosis had deteriorated rapidly during the period of usage of the first prosthesis, and she expired soon after that. The youngest subject seen, weighing less than 90 pounds, and standing less than five feet tall, could not be provided with the Otto Bock hip joint as this was too bulky and too heavy.

An inexpensive hinge was designed for this patient. Consisting of a piano-type hinge, it was attached distally to a PVC plug designed for attachment of the SACH foot to the PVC pylon. In this case, however, the PVC plug is placed upside down in the proximal end of the pylon. The distal portion of the hinge is attached to this plug and the proximal portion to the socket. Advantages of this particular hip joint design are low weight, low cost, ready availability, and ease of application. This design was subsequently used in four additional pediatric and adult patients.

As patients progressed to a new socket with a hip joint, most regarded it as a welcome change. Except for one patient who wore the preparatory prosthesis six to eight hours a day, usage in the home was limited with the non-articulated pylon (first phase prosthesis). One of the older patients reported using it with a walker in the home for moderate distances. Some of the pediatric patients wore it for a half hour to one hour at a time a few times a day. The reasons for such limited use were mainly attributed to difficulties in sitting.

Two patients developed wound infections early in their gait training, and had to discontinue wearing the prosthesis until they were medically cleared. Lack of usage by one of the pediatric patients was the result of poor cooperation on the part of the parents. In this case, many communication difficulties existed, with instructions frequently misunderstood. Another patient became overwhelmed by drastic personal and family problems after her participation had started. She cancelled or missed several gait training sessions. After an extended period of time, she was discontinued from the project.

The second phase prosthesis with pylon and articulated hip joint allowed the subjects to sit more comfortably. Most of them learned to initiate hip flexion for the swing phase and to control extension in the stance phase during the first physical therapy session. A few required cueing and reinforcement to fully weight-shift over the prosthetic leg during the stance phase to attain a position of extension. After more practice sessions, this movement became increasingly more natural. Less exaggerated and forceful pelvic tilt was needed to initiate swing phase than with the non-articulated Phase one prosthesis. This made ambulation smoother and easier for some patients. The increased weight and bulk of the hip joint and additional surrounding materials did not seem to cause concern for any of the patients. Transition to use of the heavier prosthesis is believed to have been eased by training with the lighter, simpler, and safer Phase one preparatory prosthesis.

The second phase prosthesis was used by the majority of the subjects functionally, indoors and outdoors. Wearing time ranged from three hours to 14 hours a day, the older patients using walkers, and the younger patients, forearm crutches.

As with the Phase one prosthesis, there was some dissatisfaction with certain aspects of the prosthesis. Although general mobility had definitely improved, walking speed was slower with the prosthesis than without it, i.e., with one leg and crutches. This particularly limited function for one of the younger patients who spent much of his time outdoors. One of the adult patients reported he was unable to get in and out of his car when wearing the prosthesis. Understandably, many patients were not pleased with the cosmesis of the temporary prosthesis. Even though they still consented to participate, they felt that the straight plastic

pylon without a knee joint or foam cover was crude and socially unacceptable.

Generally, there were no mechanical problems with either of the preparatory prosthesis designs, except with some of the earlier Aquaplast material. Of the total number of sockets molded (at least two per patient), three Aquaplast sockets eventually cracked. According to the manufacturer, some of the material had a limited shelf-life, e.g., three months, after which it tended to crystallize. This was rectified in later production runs.

DEFINITIVE PROSTHESIS

Biomechanical Analysis

Hip Disarticulation

The ideal hip disarticulation socket for weight bearing should have as much horizontal area as possible. Conversely, the ideal socket for swing phase should have as much vertical area as possible to minimize "pistoning." This conflict has been empirically reconciled in a socket of diagonal shape with an upward concavity as viewed in the frontal plane (Fig.11).

When one considers the human pelvis as seen in Figure 12, Vectors A, B, and C indicate areas that most effectively receive weight-bearing forces from the socket. Vector D could lessen pistoning. These might be termed primary vectors in that they directly affect the functioning of a prosthesis during midstance and during the swing phase. All horizontal vectors might be termed secondary in that they serve a stability role—preventing the prosthesis from moving away from the pelvis in the anterio-posterior and medio-lateral directions. Horizontal vectors will only be cancelled by equal and opposite vectors on the contralateral side. This results in a "squeezing" of the patient, and discomfort. It is, therefore, desirable to maximize the primary and minimize the secondary vectors.

Early socket design was fundamentally in accord with the above, but full advantage was not taken horizontal surfaces of the pelvis. Basically, these sockets took a square form (Fig. 13). The ischial tuberosity was the primary transmitter of weight-bearing loads to the pelvis. Soft tissue was left to support indirectly

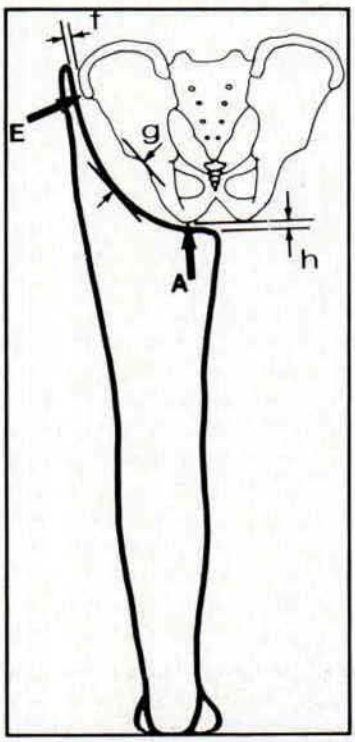

Fig. 11. Transverse cross-section of part of a conventional hip-disarticulation socket.

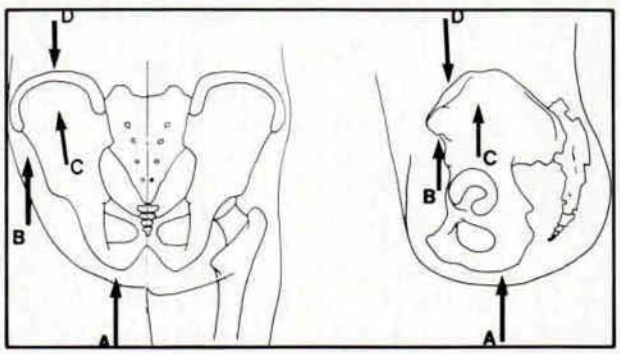

Fig. 12. Schematic of hip-disarticulation showing areas that bear weight most satisfactorily (A,B,C) and an area useful in controlling piston action (D).

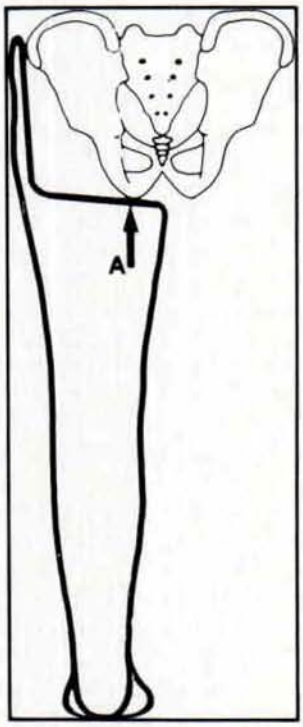

Fig. 13. An early design of socket for the hip-disarticulation case. The ischial tuberosity bore most of the downward vertical load.

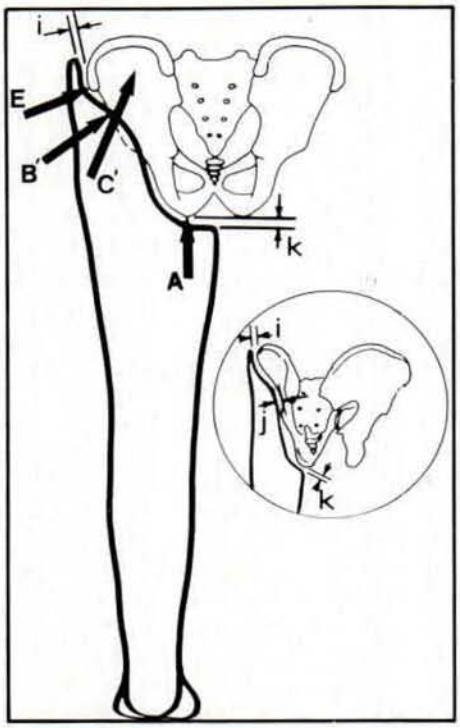

Fig. 14. "W" socket designed to spread the weight-bearing load on a hip-disarticulation stump.

the remainder of the pelvis. This was followed by a design which is currently accepted as standard and basically has a diagonal form (Fig. 11). More of the pelvis is thus directly supported.

It seems logical to extrapolate to a socket which has a "w" shape (Fig 14), in which even more of the pelvis is supported directly. Superimposition of the above figures (Fig. 15) emphasizes the regularity of the evolution.

Looking more closely at the relationship between the socket and the pelvis in figures 11 and 14, the great disparity among theoretical distances f, g, and h, as compared to that among distances i, j, k, will be noted. This is a strong indicator of more efficient use of the pelvis in bearing weight more directly rather than "floating" in soft tissue which results in pistoning while the horizontal or weight-bearing surfaces of the pelvis are not used to their fullest potential. Also, the tendency of soft tissue to flow away from the central region to the bony regions should tend to "pad" the latter for better stress distribution.

Hemipelvectomy

Establishing a socket design to transmit primary (weight-bearing) loads to the body in the case of hemipelvectomy can be approached as has been done with hip disarticulation, i.e., by considering that horizontal surfaces are best for weight bearing (Fig. 16). Theoretical primary weight-bearing areas are indicated by vectors A, B, and C.

A theoretical proposition can follow. Let vector D of Fig. 17 be the vertical component of all the forces transmitted to the body by the proposed socket (less B and C). Vector E is drawn for comparison; it represents the vertical component of all the forces transmitted to the body by a conventional socket (less B and C). Note that D, if extended upward, would pass closer to the center of gravity of the body than E would.

Basically this can be accomplished by wrapping the pelvis tightly in the vertical cross-hatched area of Fig. 17, and loosely in the horizontal cross-hatch area when taking a cast. Also, this increases the density of the tissue in

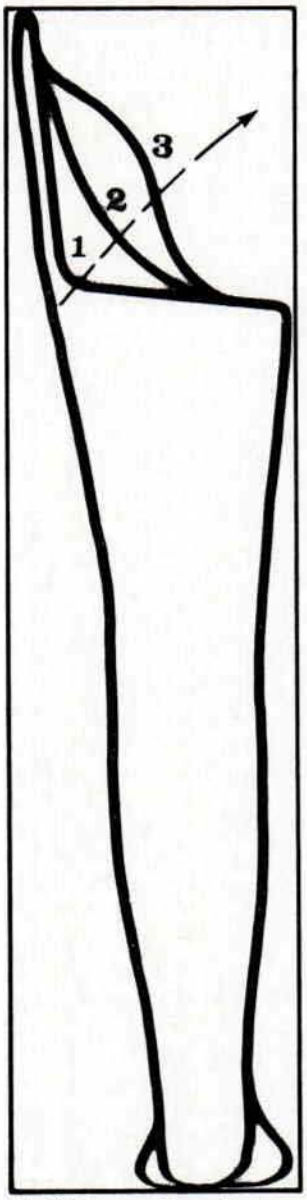

Fig. 15. Comparison of various sockets with respect to area of weight-bearing. 1, "square" socket of early design; 2, concave socket, which represents current practice; 3, "w" socket designed to spread the weight-bearing load even farther than is possible with design 2.

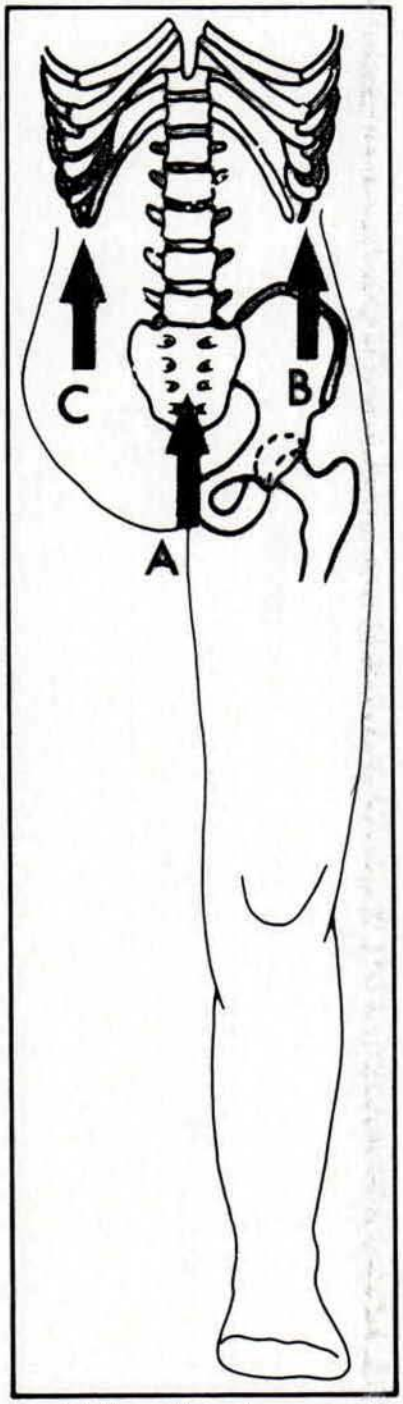

Fig. 16. Theoretical primary areas of weight-bearing for the hempelvectomy prosthesis, A, B, and C.

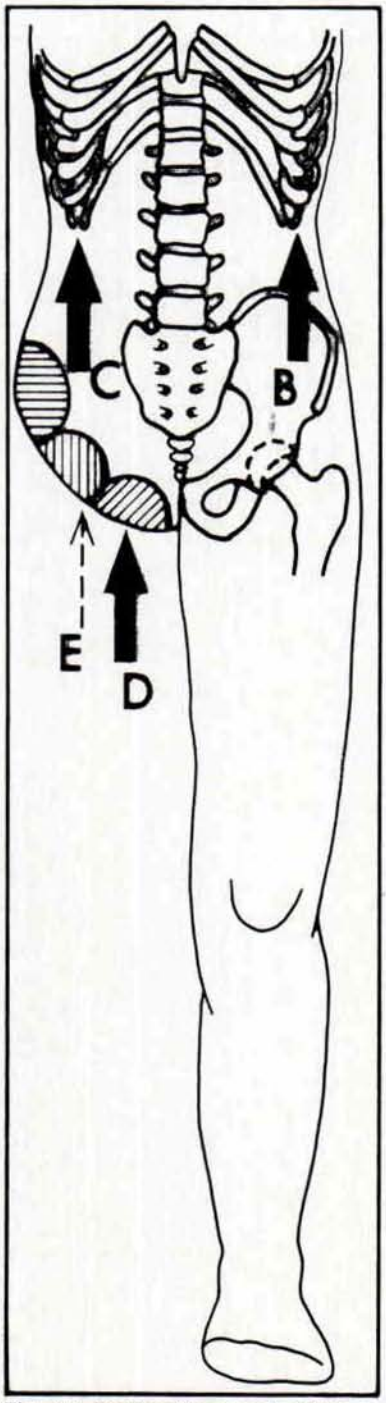

Fig. 17. Vector diagram showing application of forces provided by the new socket design. "D" represents the vertical component of all the forces transmitted by the socket, less "B" and "C", in the conventional design, and is shown here for the purpose of comparison.

the diagonal cross-hatched area which will help to keep the patient from sinking into the prosthesis upon weight-bearing and from stretching the perineum, and thus preventing a common cause of discomfort. This area of increased density is more effective in bearing weight than the other areas, indicating that weight-bearing forces are shifted medially and closer to the center of gravity.

The dashed line of Fig. 18 indicates the shape of the proposed socket in the transverse plane.

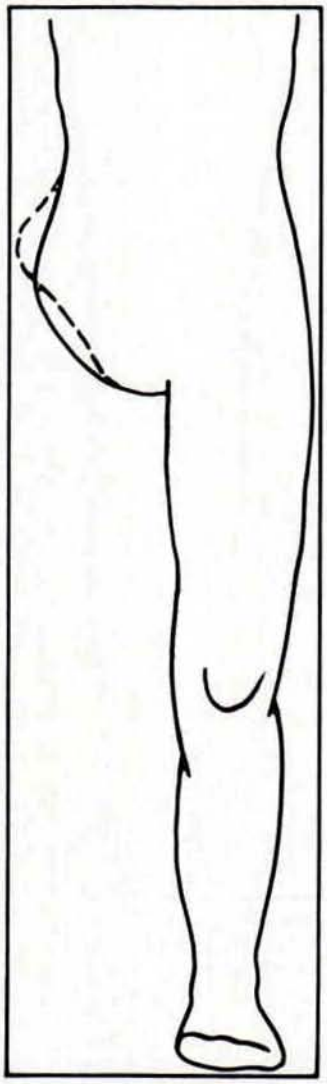

Fig. 18. Transverse cross-section of hemipelvectomy patient. The dotted line represents contour of the proposed socket at the lateral aspect.

SOCKET-FORMING APPARATUS

A socket-forming apparatus in compliance with the biomechanical analyses described above was designed to impart the appropriate biomechanical configuration onto the plaster negative cast to be taken of the patient's pelvis. The apparatus consists of half of a commercially-available plastic molded pelvic girdle used in spinal orthotics, fitted over the contralateral side (Fig. 19). Posteriorly, a flexible polyester laminated section is pivotally attached on the inside of the rigid half shell (Fig. 20). It is adjustable along a diagonal line to accommodate various sizes. This flexible section runs from its attachment posteriorly diagonally under the amputation site and is adjustably attached by means of a Velcro strap to the anterior portion of the rigid half shell. A flexible rubber tubing is attached to the posterior superior corner of the half shell by means of a parachute cord and runs anteriorly over the crest of the ilium to an adjustable attachment clip on the anterior-superior portion of the half shell.

This apparatus may be used not only for hip disarticulation amputees but for complete and modified hemipelvectomy amputees as well. Prior to wrapping the pelvis with plaster bandages, the apparatus is adjusted to the patient's pelvis (Fig. 21). The most important aspect of this steps is to place the flexible section at the appropriate angle as described in the biomechanical analysis around the amputation site and adjusting it as tightly as possible to achieve an appropriate degree of compression. After, the adjustment settings are marked and the apparatus removed, the plaster cast is wrapped and the socket-forming apparatus applied over the wet plaster cast (Fig. 22). It is adjusted in such a way that the adjustment markings are approached within a 1 cm allowance for the thickness of the plaster cast. The plaster is allowed to harden while the apparatus imparts the particular biomechanical configuration. Following this procedure, the negative cast represents very closely the actual fit of the prosthetic socket. Thus, the positive cast requires only minor modifications, involving relief build-ups for the crest of the ilium on the

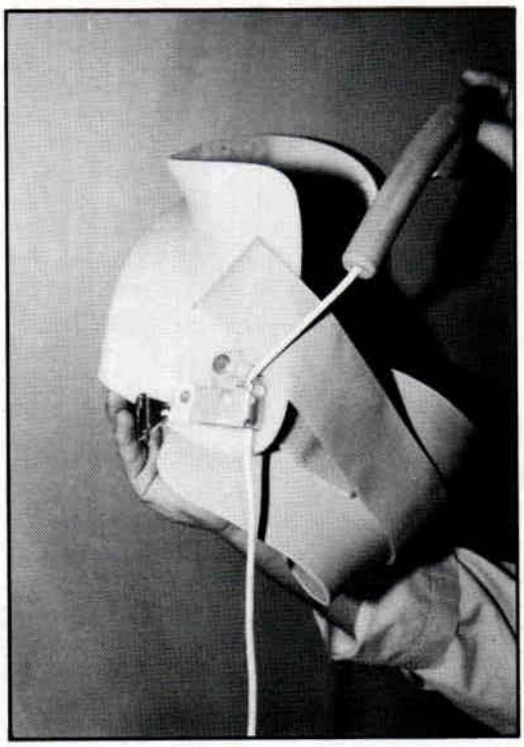

Fig. 19. Apparatus for forming the socket of cast for hip-disarticulation and hemipelvectomy patients.

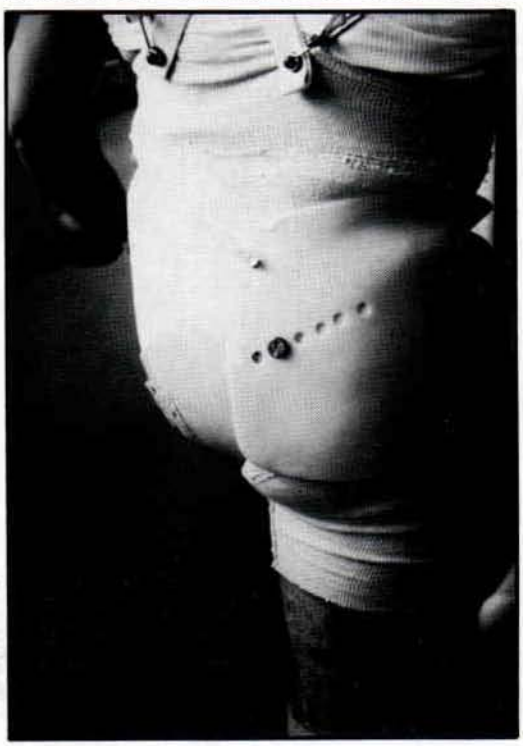

Fig. 20. The socket-forming apparatus in place on patient.

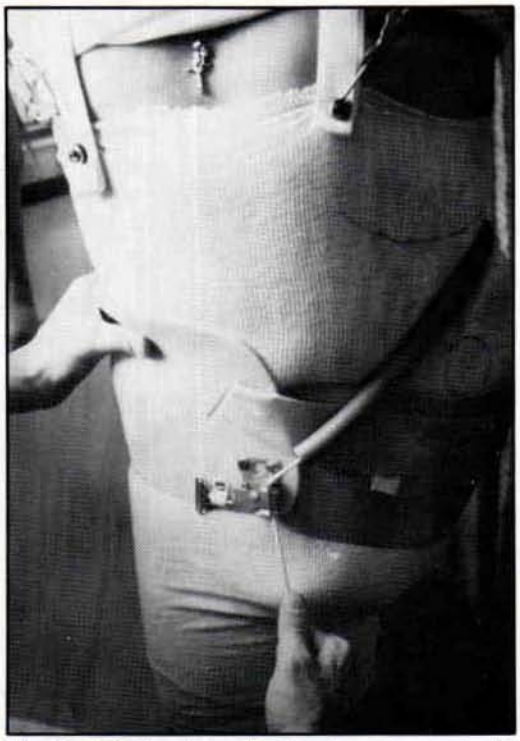

Fig. 21. Adjusting the socket-forming apparatus on the patient prior to application of plaster.

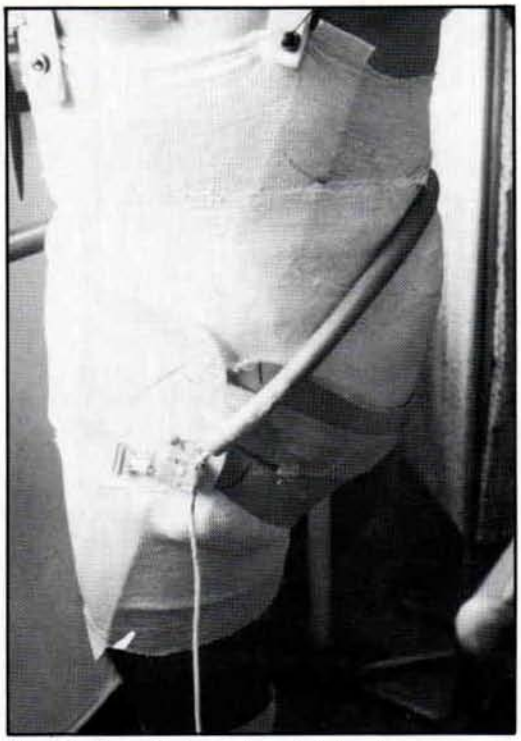

Fig. 22. Lateral view of socket-forming apparatus applied over a plaster wrap during curing stage.

ipsilateral side only.

As mentioned above, the same apparatus is used in casting a hemipelvectomy amputee, although there is no ilium on the ipsilateral side. The procedure is identical, with the rubber tubing in this case placed at the same level as the contralateral side. The reason for this is to achieve a more cosmetic appearance on the involved side, as well as to enhance suspension through the indentation created by the rubber tubing.

Biarticular Control System

Design

While one of the major purposes of the project was to design cost-effective systems for early fitting, described above, another purpose was to advance the state of the art of definitive prostheses by addressing current problems in the control of prosthetic hip and knee joint movement. The conventional hip disarticulation or hemipelvectomy prosthesis may best be viewed as a compound pendulum with practically no coordinated control of hip and knee movement. Whatever control that does exist, is by means of elastic straps, rubber bands, or springs. This results in the patient being limited to one cadence only; e.g., if the patient attempts to walk faster, joint excursion at the hip and knee is increased, which results in unequal step length and timing. Stability in the stance phase is totally dependent on alignment stability. To overcome these problems, a Hydra-pneumatic cylinder[5] was adapted to an Otto Bock modular system to coordinate hip-knee movement in a "synergistic" way (Fig. 23). This results in greatly improved cadence response of the system due to its hydraulic characteristics, as well as increased stability in the stance phase. The location of attachment points of the Hydra-pneumatic cylinder are critical to proper function of the system. Based on functional analysis, a 6:1 ratio of hip to knee flexion resistance was found to produce optimal hip-knee control. The proximal attachment is located anterior and superior to the center of the hip joint. The distal attachment is located posterior and distal to the knee center. Both the proximal and distal attachments are mounted on adjustable brackets to accommodate for static and

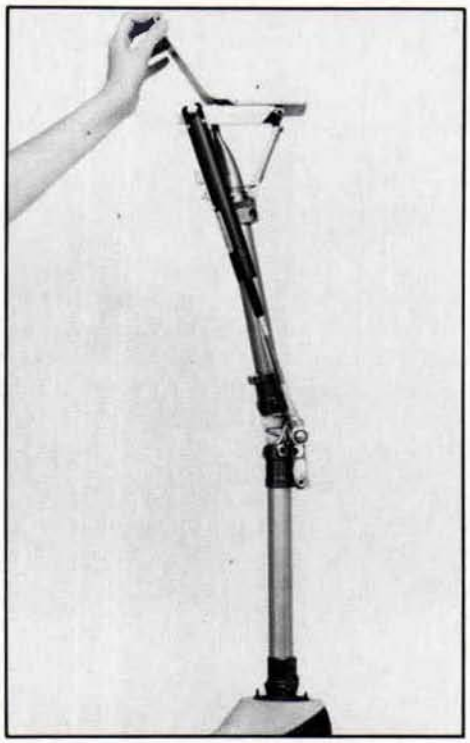

Fig. 23. Lateral view of hip-disarticulation prosthesis in which a hydraulic swing-phase control knee unit is used to control the hip.

dynamic alignment changes.

During early swing phase on the prosthetic side, fluid displacement in the cylinder dampens hip and knee flexion (Fig. 24), while a spring and pneumatic chamber in the cylinder are compressed. In late swing phase, the spring recoils while the pneumatic chamber expands, resulting in hip and knee extension.

Patient fittings

A total of three patients received the definitive prosthesis incorporating the biarticular control system. The first patient, D.S., had had a hip disarticulation 16 months before referral to the project. She had been using a hip disarticulation prosthesis with a spring-loaded hip flexion bias for one year. Her complaints with the prosthesis were knee instability induced by the hip flexion bias and a circumducted gait.

The other two patients, K.M. and R.S., were fitted approximately five months postsurgery. They had previously participated in the project and were provided the series of preparatory

Fig. 24. Lateral view of prosthesis shown in Figure 23 being used by a patient.

prostheses. K.M. was a modified hemipelvectomy amputee who had been using a conventional endoskeletal prosthesis for three months. She was unhappy with her present gait pattern which showed severe circumduction and lack of knee flexion in swing. She expressed a strong

desire to receive a new prosthesis with biarticular control. R.S. was a regular hemipelvectomy amputee who was still using her preparatory prosthesis and who in the past had medical complications which had delayed fitting of the definitive prosthesis.

Physical therapy instruction and training does not differ with this system compared to training with a conventional endoskeletal Canadian-type hip disarticulation prosthesis. What is unique is the monitoring and fine tuning possible with the resistance adjustment of the Hydra-pneumatic unit as well as the mechanical friction of the knee to achieve a smooth, coordinated gait pattern.

Initially, the mechanical knee friction should be decreased to zero while the hydraulic resistance is increased. If lack of training or bad habits cannot justify a patient's difficulty in tilting the pelvis posteriorly to initiate swing, the hydraulic resistance should be decreased slightly. Proper step length may have to be sacrificed initially, but once the tilt improves, hydraulic resistance can be increased again.

When heel rise is too high, resulting in excessive length of time for the prosthesis to extend to the weight-bearing position, the following adjustments are necessary.

If this deviation occurs at the beginning of training, the Hydra-pneumatic resistance should be decreased slightly and the mechanical resistance increased slightly. When the gait pattern improves, resistance in the hydraulic unit once again is increased and the mechanical friction decreased. Both types of resistance may need to be adjusted alternately until the condition for optimal swing is found.

Since D.S. and K.M. had already ambulated with permanent prostheses, much of the gait training involved relearning a proper gait pattern. This was believed to be markedly enhanced with the "synergistic" control system on the basis of clinical observation and patient comments.

The last patient, R.S., who had not yet been using a permanent prosthesis, was the ideal candidate for this project. She had mastered basic body mechanics with the temporary device, and had not developed any negative habits in her gait pattern. She progressed rapidly in her gait training and completed it within one month.

While other patients participating in this project progressed to a definitive prosthesis, they were fitted prior to the development of the biarticular control system, i.e., with conventional systems, and thus are not described in any detail here. Nevertheless, all five amputees advancing to the definitive fitting stage were provided with sockets made through the use of the socket-forming apparatus. The only difference amongst the sockets was the location of the opening. The more recent sockets were fitted with a posterior opening, which is believed to be more effective for weight-bearing purposes since better advantage can be taken of increased intra-abdominal pressure with a solid anterior socket. Furthermore, any volume changes are better accommodated with a posterior opening since the crucial skeletal relationships, e.g., distance between the anterior superior spines, are not affected in this way.

Acknowledgements

Much of this research was made possible through the fine cooperation of the Surgical, Rehabilitation, Medical, and Physical Therapy Departments at Memorial Hospital, with which an official research affiliation existed during the latter part of this project.

The contributions of Mr. Robert Wilson, M.S., and Mr. Adam Kubert, B.A., for the fine art work, and Mr. Stan Willard for photography are gratefully acknowledged.

Last, but not least, the project director, wishes to thank all project personnel, and especially the patients, for their dedicated participation.

Footnotes

[1] This investigation was supported, in part, by Research Grant Number CA 20703-03 from the National Cancer Institute, National Institutes of Health, Department of Health, Education, and Welfare, Bethesda, Maryland 20014.

[2] Institute of Rehabilitation Medicine, New York University Medical Center, 400 East 34th Street, New York, NY 10016.

[3] WFR Corporation, Ramsey, New Jersey.

[4] Otto Bock Orthopedic Industry, Inc., 610 Industrial Avenue, Minneapolis, Minnesota 55422.

[5] Kingsley Manufacturing Company, 1984 Placentia Avenue, Costa Mesa, California 92627.

Orthotics and Prosthetics, Vol. 35, No. 2, pp. 29-32, June 1981

Some Thoughts About Nomenclature For Limb Prosthetics[1]

CHARLES H. PRITHAM, C.P.O.[2]

In a document (3) published in 1974, Hector W. Kay reported on decisions made by the Task Force on Standardization of Prosthetic-Orthotic Terminology of the Committee on Prothetic-Orthotic Education of the National Academy of Sciences relating to nomenclature for limb prosthetics. The gist of this report was that the currently accepted system of "Americanisms" should be abandoned in favor of an international terminology; specifically that amputation levels and corresponding prostheses should be named in the same fashion as the then recently developed technique for naming transverse congenital deficiencies that essentially "present as amputation-like stumps" (1). In effect, by bowing to the necessity of comparing the new congenital terminology back to the (presumably unacceptable) standard terminology the Task Force nullified the argument in favor of the change. If the situation currently prevailing is all that clear then why change it?

The Task Force acknowledged this point explicitly on pages 38-39 by stating that prosthetics nomenclature was much clearer than orthotics nomenclature, and therefore, there was less urgency in modifying it. The most telling argument against the prevailing situation was that amputations at or about the same level were referred to by different names: "knee disarticulation" versus "exarticulation" versus "Gritti-Stokes." In considering the other points made in the introduction, it is difficult to see how they apply to prosthetics nomenclature to any degree approaching the same fashion that they apply to the confused state of orthotic nomenclature that prevailed at that time. That the orthotic nomenclature in use needed urgent overhaul is undisputed, but to argue by inference that similar drastic efforts should be made in prosthetics is invalid and completely overlooks the confusion that is likely to result if drastic changes are introduced. Events since 1974 seem to bear out this point. The revised orthotics nomenclature has been adopted wholeheartedly, precisely because it fills a pressing need; no such ground swell of popular support has arisen for modification of prosthetic nomenclature. A similar situation undoubtedly exists in the classification of congenital limb deficiencies. The system for classifying longitudinal deficiencies has been accepted at least in part because it clearly fulfills a need. To the best of the author's recollection he has never heard anyone refer to a transverse deficiency in terms of the system; rather the phrase "congenital BK" (or whatever) is far more likely to be used.

It may be taken then that no argument exists favoring dropping of the current system of prosthetic nomenclature. It may also be stated that no argument exists for adopting the proposed new nomenclature; quite the opposite view. Change for the sake of change, and without much prospect for clear, overwhelming benefits to accrue is illogical, especially when considered in light of the confusion and resentment it is likely to engender.

That confusion is apt to result strikes me as all too likely. To name what remains by specifically citing what was removed is more than a lit-

tle perplexing. The use of the words leg, thigh, arm, and forearm as they are used in the proposed terminology are totally foreign to most Americans; when Americans say "leg" they mean from the hip down, not from the knee down. While it is all well and good to avoid fobbing "Americanisms" off on the rest of the world, any new terminology adopted should not represent a radical departure from the old since a very large proportion of the modern prosthetic literature is written in terms of current usage. The proposed system is not at all elegant with its system of trailing modifiers, nor does it readily lend itself to abbreviation and acronyms. Compare the simplicity and tidiness of "AFO" or "BK" with "leg, partial (middle ⅓)."

The situation in which any method of classifying congenital limb deficiencies is employed is not at all comparable to the situation in which a system is meant to be used by highly specialized individuals about a very complex, although small, group of patients. Not only is clarity of communication of vast importance, but a high degree of motivation is needed for mastering a complicated vocabulary. The world of "acquired amputation" is not nearly so exclusive a domain. A vast population of personnel, widely differing in educational level and motivation (surgeons, prosthetists, therapist, engineers, case workers, clerical help, administrators, bureaucrats, etc.) must communicate to each other in terms that are mutually acceptable. Furthermore, the two causes of limb loss are quite different in terms of prognosis and attending complications (phantom pain/sensation, neuromas, bony overgrowth to name but a few). To lump congenital problems and acquired amputations together, and gloss over the distinctions, is to do someone a disservice.

Where do we stand then? The need is for a clear, explicit system of nomenclature that is readily applicable to both amputations and prostheses and can be simply translated. It should be written in terms of well defined, distinct anatomical sites and make distinctions only where clearly distinguishable functional implications exist. A joint (or to be even more precise a joint line) is a clear and well defined anatomical unit, whereas reference to one of the limb segments is vague and must be qualified (mid-thigh, distal humerus, etc.). Reference to a "below-knee amputation" or "above-elbow amputation" is not altogether illogical then. It is interesting that we should so readily accept the method as applied to orthotics (AFO, CTLSO) yet stand ready to abandon it it reference to prosthetics. I still feel that a case can be made for referring to an AFO as a below-knee orthosis (BKO) (as they still do at NYU) and that prosthetic and orthotic terminology be brought into total harmony.

One useful point from the 1974 report of the Task Force, and that is the distinction of amputation versus disarticulation. Amputation (besides its wider connotation encompassing the entire field) refers to the severing of a limb through the shaft of diaphysis. It was intended by the task force that disarticulation should refer not only to severance through the joint space (without the cutting of bone) but also to those instances when the remaining long bone was cut distal to the distal epiphyseal plate. This argument may be extended by inference to include all amputations through the distal condyles and thus encompass all procedures such as the "Gritti-Strokes." Functionally there is little to distinguish a true knee disarticulation from those amputations performed through the condyles and the prostheses are essentially identical.

The quest to eliminate eponyms means that a Syme's amputation should be referred to as an ankle disarticulation. This may well be the case but if ever there is an exception that proves the rule this is it. The Syme's procedure is seductively appealing but oftentimes performed with discouraging results. Apparently considerable skill and attention to detail are necessary to preserve the heel pad and keep the attendant blood vessels intact, and to cut the tibia in the proper plane; nor are all patients with peripheral vascular disease who might otherwise benefit by it suitable candidates. Perhaps it is then that the eponym should be retained in tribute to this great Scottish surgeon and in warning to those who might seek to emulate his example.

The Task Force recommended that amputations through the long bones (BK, AK, BE, AE,)

should further specify through which third the amputation was performed. Earlier it was stated that distinctions should be made only when there are clearly distinguishable functional implications. This is intended for the sake of brevity. There is only one level where such a further breakdown may be considered truly useful and that is the below-elbow (residual prona tion/supination is a function of residual length). Otherwise it is hard to draw inferences about amputee performance as affected by stump length nor is it particularly easy to relate changes in socket shape and prosthetic prescription to changes in length. A clear functional distinction exists between a knee-disarticulation socket although both have more or less quadrilateral brims. It is far more difficult to distinguish differences between a socket for a long above-knee stump and short one. If it is considered desirable to provide any indication about stump length then it should probably be given as a percentage of the bone length on the sound side. This is far more precise and is readily possible with a few simple measurements and calculations. Indeed, with today's proliferation of hand-held calculators it should provide clinicians with many hours of innocent fun. Partial foot and partial hand amputations should be left out of the classification scheme, as are hand and foot orthoses in the orthotic system of nomenclature.

At the meeting of the Working Group 1 it was decided that prostheses (St. Andrews, Scotland, April 1980) orthoses should be referred to as devices or systems and not as components. To me the proper word is device; to refer to a molded plastic ankle-foot orthosis with Velcro strap as a system is surely a case of verbal inflation. Webster's Collegiate Dictionary defines a "System" as a regularly interacting or interdependent group of items forming a unified whole. It defines "device" as a piece of equipment or mechanism designed to serve a special purpose or perform a special function. An example familiar to all is a stereo system. Essentially such a system consists of a signal source (turntable, recorder, or receiver)an amplifier, and transducers (loudspeakers or headphones); each is complete and self contained but helpless without the other two elements. In a like fashion the totality of patient, prosthesis/orthosis, and peripheral elements (shoes, canes, etc.) forms a system of interdependent elements. To me the use of the word "system" to refer to a prosthesis or orthosis, however complicated, is another example of creeping "computerese" or "technicalese" (as is "interface") that must be avoided. Do patients access their sockets or input their stumps? Does a decubitus ulcer constitute impact?

To reiterate then a case is made for preserving the present terminology (Table 1) and for not adopting the system for classifying transverse congenital deficiencies. It is suggested that prostheses and orthoses be referred to as "devices", not as "systems".

Table 1

Partial Food (P.F.	Partial Hand (P.H.)
Symes (S)	Wrist Disarticulation (W.D.)
Below Knee (B.K.)	Below Elbow (B.E.)
Knee Disarticulation (K.D.)	Elbow Disarticulation (E.D.)
Above Knee (A.K.)	Above Elbow (A.E.)
Hip Disarticulation (H.D.)	Shoulder Disarticulation (S.D.)
Hemipelvectomy (H.P.)	Forequarter (Fq.)

Hemicorporectomy (H.C.)

Abbreviations would be for prosthesis (P) or amputation (A) to construct appropriate acronyms (B.K.P., B.K.A.)

Table 2

	American Usage		French	German	Russian?	Chinese?	Spanish?	Arabic?
Amputation	Amputation							
Disarticulation	Disarticulation	A						
Foot	Foot	D						
Ankle	Ankle	A						
Knee	Knee	K						
Hip	Hip	H						
Hand	Hand	H						
Wrist	Wrist	W						
Elbow	Elbow	E						
Shoulder	Shoulder	S						
Above	Above	A						
Below	Below	B						
Partial	Partial	P						

Footnotes

[1] Branch Manager, Snell's is of Louisville/Division of Durr-Fillauer Medical, Inc. 744 E. Broadway Louisville, KY 40202

[2] This document was prepared for use by the working Group I (Classification & Nomenclature) of Technical Committee 168 (Prosthetics & Orthotics) of the International Standards Organizations

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(1) Harris, E.E., A new orthotics terminology; a guide to its use for prescription and fee schedules, Orthotics and Prosthetics. 27:2:6-19, June 1973.

(2) Harris, R.I., *The history and development of Syme's amputation. Selected Articles from Artificial Limbs*, pp 233-272. Robert E. Krieger Publishing Co., Inc., Huntington, New York, 1970.

(3) Kay, Hector W., *A proposed nomenclature for limb prosthetics. Orthotics and Prosthetics*, Vol. 28, No. 4, pp 37-47, Dec. 1974.

Orthotics and Prosthetics, Vol. 35, No. 2, pp. 33-37, June 1981

Technical Note

Use of a Pelite Insert with the Muenster Socket

The so-called Muenster socket has been used for many years in fitting amputees who have short and very short below-elbow stumps in order to provide suspension of the prosthesis. In our facility we have also used the Muenster socket on many long BE cases and some wrist disarticulees when the figure-eight harness and biceps cuff presented problems.

In the beginning, we found that the long BE and WD cases were unable to don the Muenster prosthesis when provided as described; that is, as a hard socket. Experimentation led to the development of a Pelite-lined Muenster socket (Fig. 1) that has proven to be quite successful for use at any level of amputation below the elbow.

Fabrication procedure is as follows:

The casting procedure is unchanged, elastic plaster-of-Paris bandages reinforced with ordinary plaster-of-Paris bandages being used.

The positive model needs to be modified on the plantar surface to make stump access possible. Figure 2 shows the area that needs buildup on a wrist disarticulee who has a measurement of 8 inches between the medial condyle and the distal end of the stump.

After modification of the positive model a Pelite insert is formed over it just as inserts are made for use in patellar-tendon-bearing below-knee sockets. A socket is laminated over the insert and model in the usual fashion. The finished insert is shown along side the socket in Figure 3, and the prosthesis with insert in place is shown in Figure 4.

At the initial fitting the proximal brim is trimmed until the patient can flex the elbow to at least 115 degrees (Figs. 5 and 6). The figure-nine harness is used (Fig. 7).

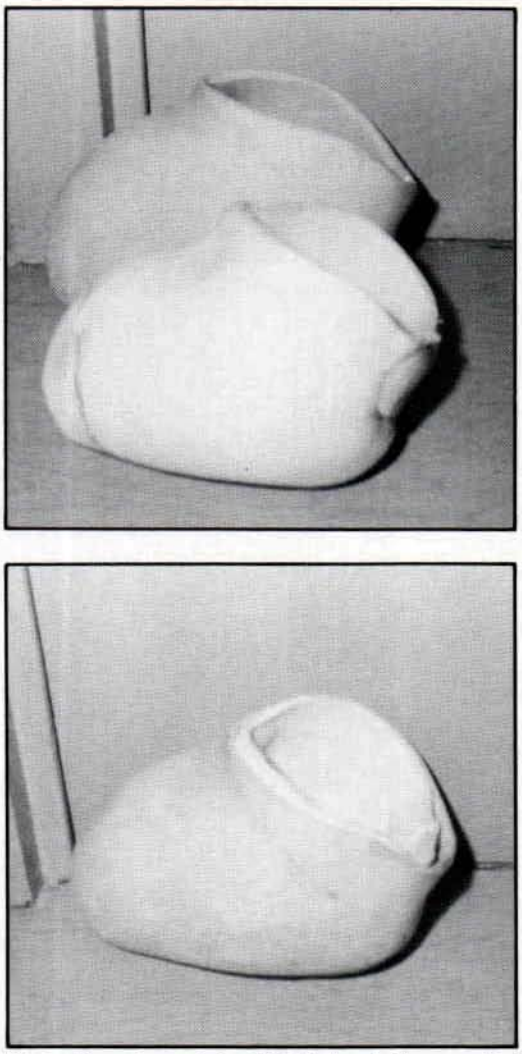

Fig. 1. Muenster socket with Pelite insert. Upper, socket and insert, side by side; lower insert in place.

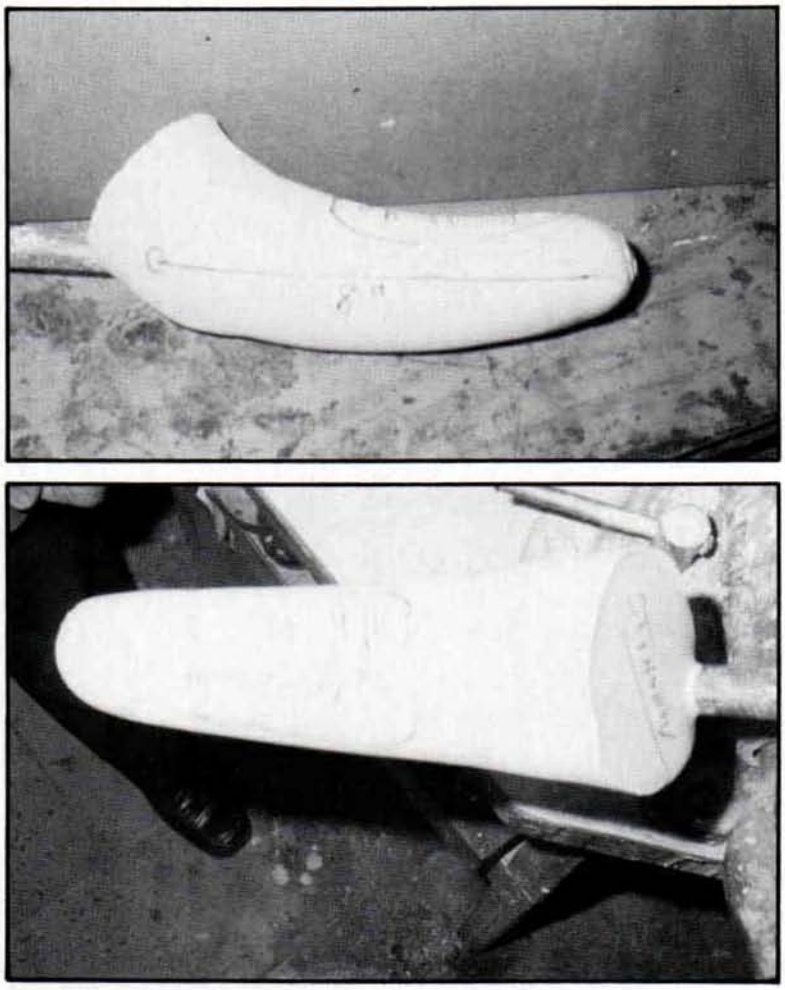

Fig. 2. Two views of positive model of wrist disarticulation stump showing location of build-up needed to provide room for insertion of the stump in the socket.

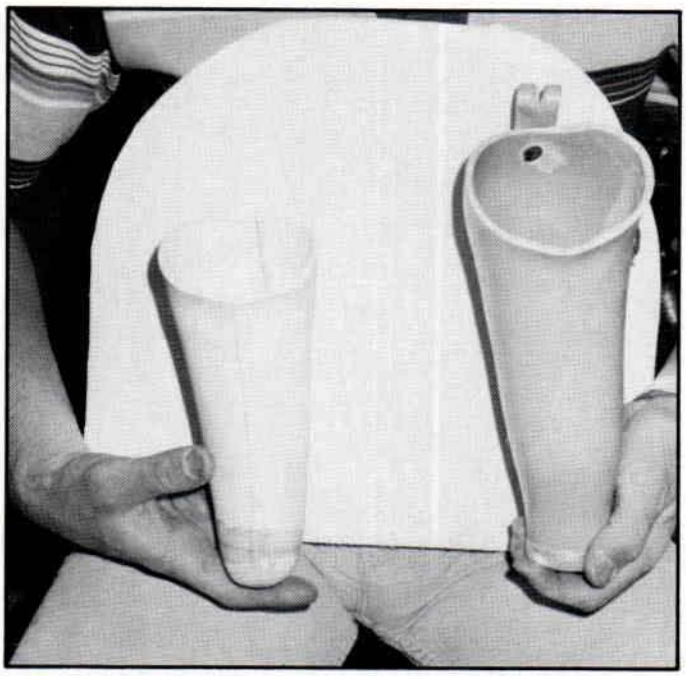

Fig. 3. The Pelite insert and the socket formed over the positive model shown in Figure 2.

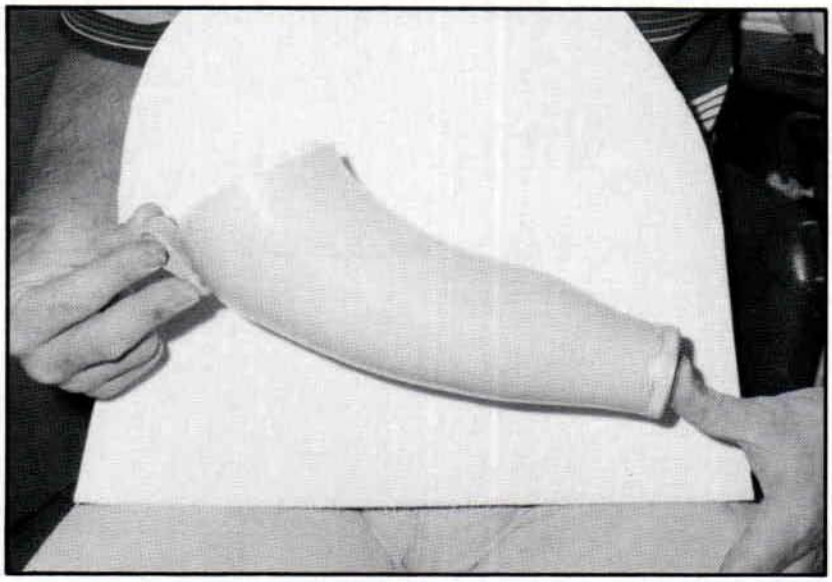

Fig. 4. The wrist-disarticulation socket with insert in place ready for initial fitting.

Fig. 5. Elbow extension during initial fitting. The amount of initial flexion built into the socket is considered to be correct.

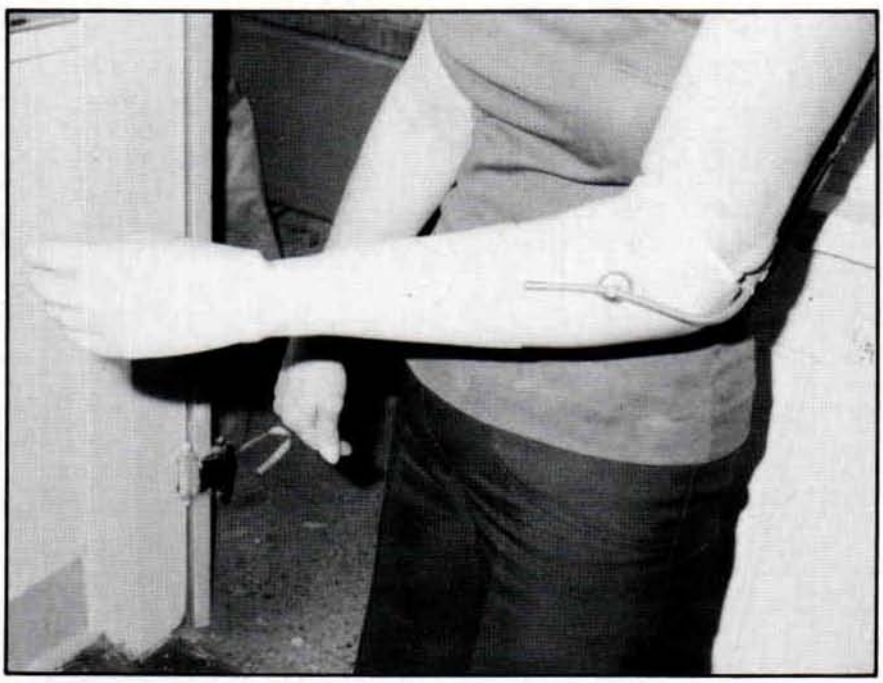

Fig. 6. Elbow flexion during initial fitting. The degree shown here is inadequate and the anterior brim must be trimmed so that at least 110 degrees can be obtained.

Fig. 7. The figure-nine harness used with self-suspending sockets.

To date we have fitted about a dozen Muenster sockets with Pelite inserts. The stumps have varied in length from 2 to 8 inches. Our experience has shown that the Pelite insert, in addition to providing additional comfort, gives the prosthetist an opportunity to make modifications for stump changes easily.

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James Cicero, L.P.T., C.P.
Hedgecock Artificial Limbs Co.
2827 Commerce Street
Dallas, TX

Reference

Kay, Hector W. et al, *The Muenster-type below-elbow socket, a fabrication technique*, Artificial Limbs, Vol. 9, No. 2, Autumn 1965.

Orthotics and Prosthetics, Vol. 35, No. 2, pp. 38, June 1981

New Publications

Human Walking by Verne T. Inman, Henry J. Ralston, and Frank Todd; William and Wilkins, Baltimore/London, 1981. 154 pp., 168 illustrations; $25.00.

When the National Artificial Limb Program was started by the National Academy of Sciences in 1945 at the request of the Surgeon General of the Army, it was the consensus of the principals involved that a good part of the then existing problems with artificial limbs could be solved by having a group of medical doctors develop criteria for the ideal artificial leg and the ideal artificial arm, and pass these along to a group of engineers, who would simply come up with designs that would meet the criteria.

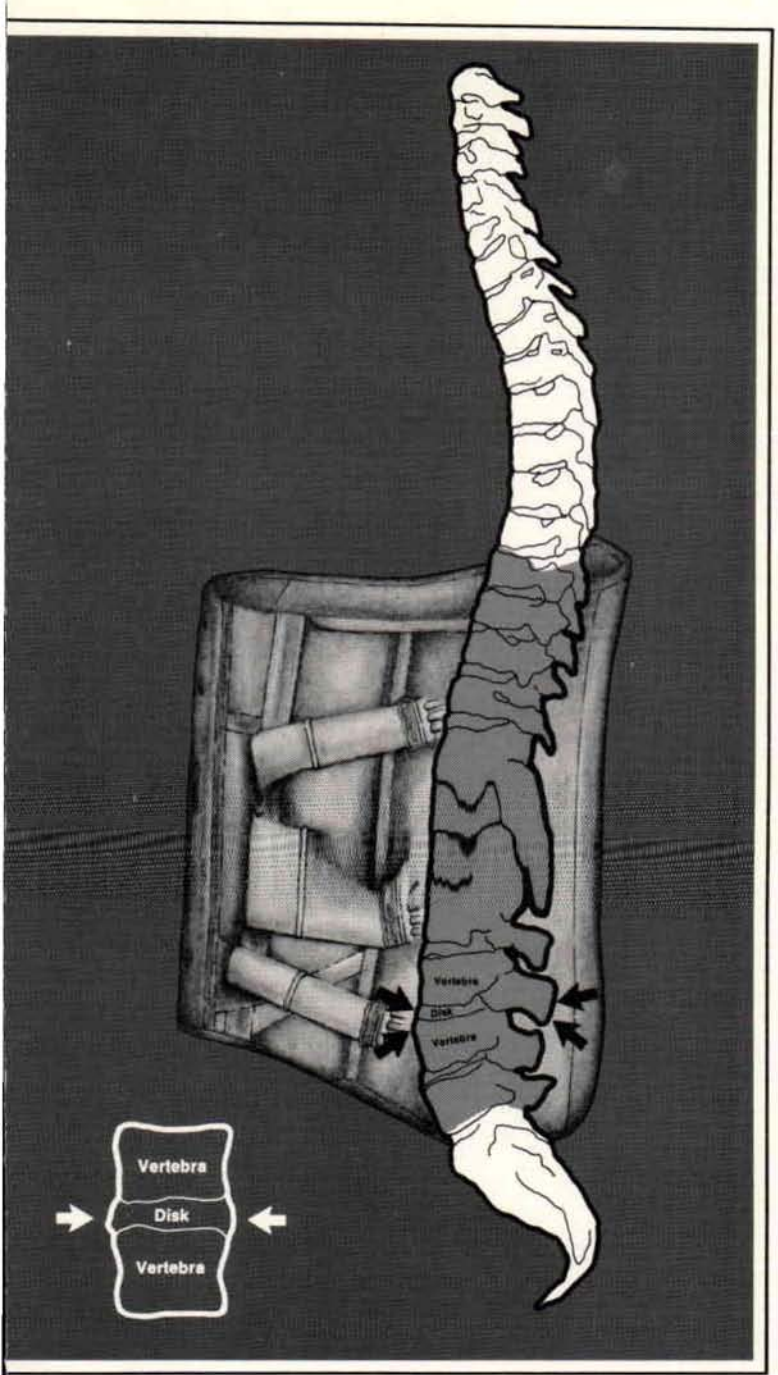
It soon became clear that the medical profession was not in a position to develop such criteria, and that if significant advances were ever to be made in artificial legs, fundamental information needed for development of design criteria had to be gained through research. To carry out this basic research the National Academy of Sciences turned to the team of Verne T. Inman, a senior member of the Department of Orthopedic Surgery, University of California, San Francisco, and Howard D. Eberhart, Professor of Civil Engineering, University of California, Berkeley, who had collaborated previously in studies concerning the function of the shoulder joint.

Using instrumentation and calculators that are crude by today's standards, the team headed by Dr. Inman and Prof. Eberhard made such excellent progress in the first three years that it was possible to develop the criteria upon which most artificial limbs developed since 1950 are based. The work has been continued to the present time. Data have been refined throughout the years and the study has been expanded to include all aspects of human walking.

Upon his retirement in 1973, Dr. Inman began several books that summarized his various works in functional anatomy. This volume is the one that covers human locomotion and application of this knowledge in the field of lower-limb prosthetics. The co-authors and other contributors were all members of the team studying human locomotion. It is most unfortunate that Dr. Inman died before publication of the book, but we are fortunate that he had completed his part before his death.

Anyone who knew Dr. Inman will know that this work is an elegant and complete treatise on the subject. Every person that is concerned with human gait, normal or pathological, should read "Human Walking" thoroughly, and have a copy of it readily available for reference in the course of their work, whether it be research or clinical practice.

A. Bennett Wilson, Jr.



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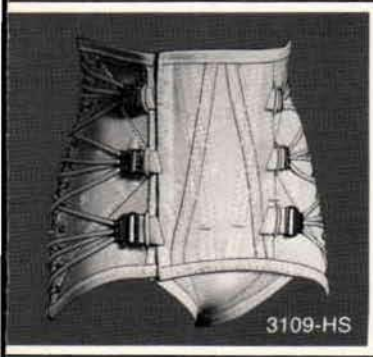

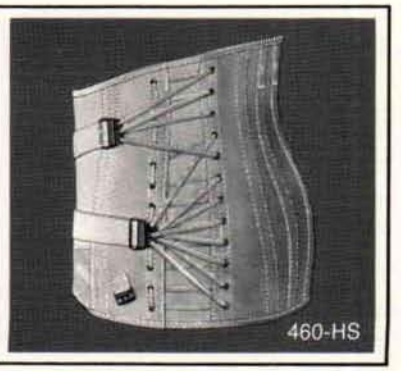

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