A NEW LIMB FOR THALIDOMIDE CHILDREN

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Summary

This paper describes a six-degree of freedom upper limb prosthesis for amelic and phocomelic thalidomide children based on a new type of rotary actuator. The energy source is compressed carbon dioxide contained in a rechargeable cylinder which may form an integral part of the limb. A new pneumatic valve, in two versions, manually and solenoid operated is described. Other features include the use of rubber springs and an optical gauge on the hook permitting the user to estimate the gripping force.

At present the limb is controlled by sequential operation of the valves by the wearer. Preliminary investigations indicate that a much simpler control method requiring only one command signal source and using a logic-gates device may be possible in the future.

Introduction

The field of powered artificial limbs research is a relatively new one. The design problems of a powered prosthesis have only recently been dealt with by engineers, partly due to the growth of interdisciplinary research and the interest aroused as a result of the thalidomide disaster.

My first encounter with the subject occurred in 1965 during a research fellowship in Oxford. A study of the gas-powered limbs then in use by thalidomide children led me to the following conclusions regarding their performance, characteristics and components [1—8]. We shall consider these successively:

1. *The actuator.* The usual type of actuator in gas powered limbs is the piston-cylinder assembly. Its disadvantages for this purpose are the large space requirements. Its diameter is greater than the effective piston area and its unextended length is about one and a half times its effective stroke.

2. *Kinematical design.* The movements at the joints of a limb are all rotations about an axis. It is therefore necessary to convert the linear motion of the piston to rotary motion about an axis with resulting complications.

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3. *Efficiency.* In the small sizes of pistons used the efficiency rarely exceeds 75 percent. The actuating linkage transforming the linear motion to rotary motion reduces to overall efficiency still further.

4. *Simulation of a normal limb.* Existing limbs are much inferior in this respect and have a smaller number of degrees of freedom. Usually, the movements provided are gripping, wrist rotation and elbow flexion with passive joints at the elbow and shoulder. The aesthetic aspect is adversely affected as a result.

5. *Forces.* The forces and moments available are much lower than those in a normal limb.

6. *Weight.* The weight is a serious problem causing chafing between the prosthesis harness or jacket and body as well as various spinal deformities due to asymmetrical loading.

7. *Control.* This is amongst the most severe problems facing the designer of artificial arms for although it is possible to attain a certain proportion of the forces and movements of a normal limb it is a different task altogether to simulate its nerve system.

Limited Objectives

Because we cannot duplicate the performance of an arm we must accept more limited objectives. We can set these out on the basis of the preceding analysis.

1. It is desirable to imitate the kinematic characteristics of the arm but the use of piston actuators severly limits design freedom. It is therefore essentail to find a more suitable actuator, preferably a rotary actuator.

2. Comparison of CO_2 as a power source with others such as electric-hydraulic or electric shows that it offers the best overall solution when the ratio of the energy available to the combined weight of container, battery reducer, piping, fluid, valves or switches and actuator are considered. It also offers much higher peak powers resulting in faster response of the limb. CO_2 will therefore be used as the energy source.

3. It is desirable to reduce the weight of the valves, return springs, and structural members.

4. It is desirable to increase the force specifications.

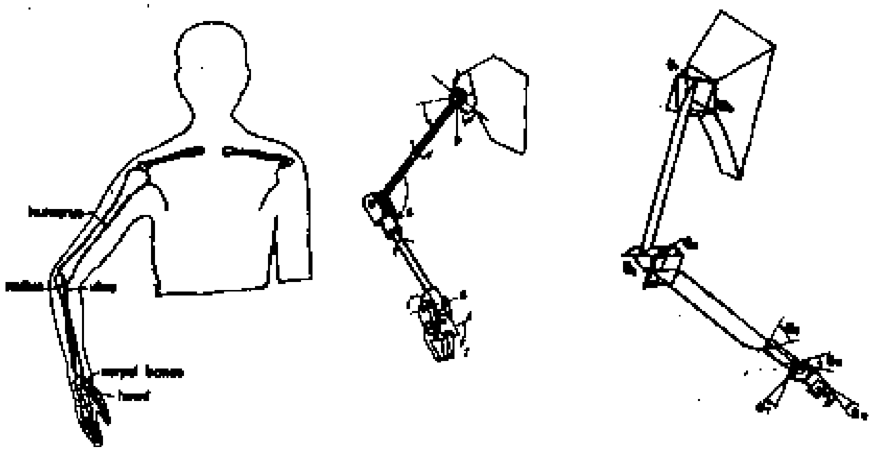
5. It is desirable to give the wearer visual or other feedback of the gripping force. The other forces such as those produced in raising an object create a reaction at the area of contact between wearer and prosthesis. The gripping force is an internal one.

6. It is desirable to simplify the gas cylinder replacement procedure and incorporate it in the limb itself for recharging by the wearer.

7. It is desirable to develop a control system which can be easily operated by patients with severe disabilities.

Kinematical and Kinetic Considerations

Considering the shoulder as a frame of reference it is seen that the hand at the end of a seven degree of freedom chain (see Figure 1a and b) shows a space mechanism kinematically equivalent to a normal arm. There are 8 degrees of freedom in all when the hand is replaced by a single degree of freedom pincer movement. Note the ball joint at the shoulder.



a) b) c)

Fig. 1.

Figure 1c shows a second space mechanism quite similar to a normal arm. This mechanism is made up entirely of hinged joints.

Taking the shoulder as the origin of our frame of reference, we see that the humerus has three degrees of freedom θ_1, θ_2, θ_3, two defining the orientation in space and one rotation about its longitudinal axis. A typical range of movements is indicated by the angle between the two extreme positions. Figure 2a shows rotation of the humerus about a vertical axis, Figure 2b shows rotation of the humerus about a horizontal axis and Figure 2c shows rotation of the fore-arm about the humeral axis.

The ulna and radius can rotate about the end of the humerus about an axis normal to the plane of the humerus and the ulna radius pair. Furthermore the ulna and radius can twist about one another thus rotating the hand about their longitudinal axis. This adds two more degrees of freedom θ_4, θ_5. Figure 2d shows elbow flexion and Figure 2e shows rotation of the hand about the axis of the forearm.

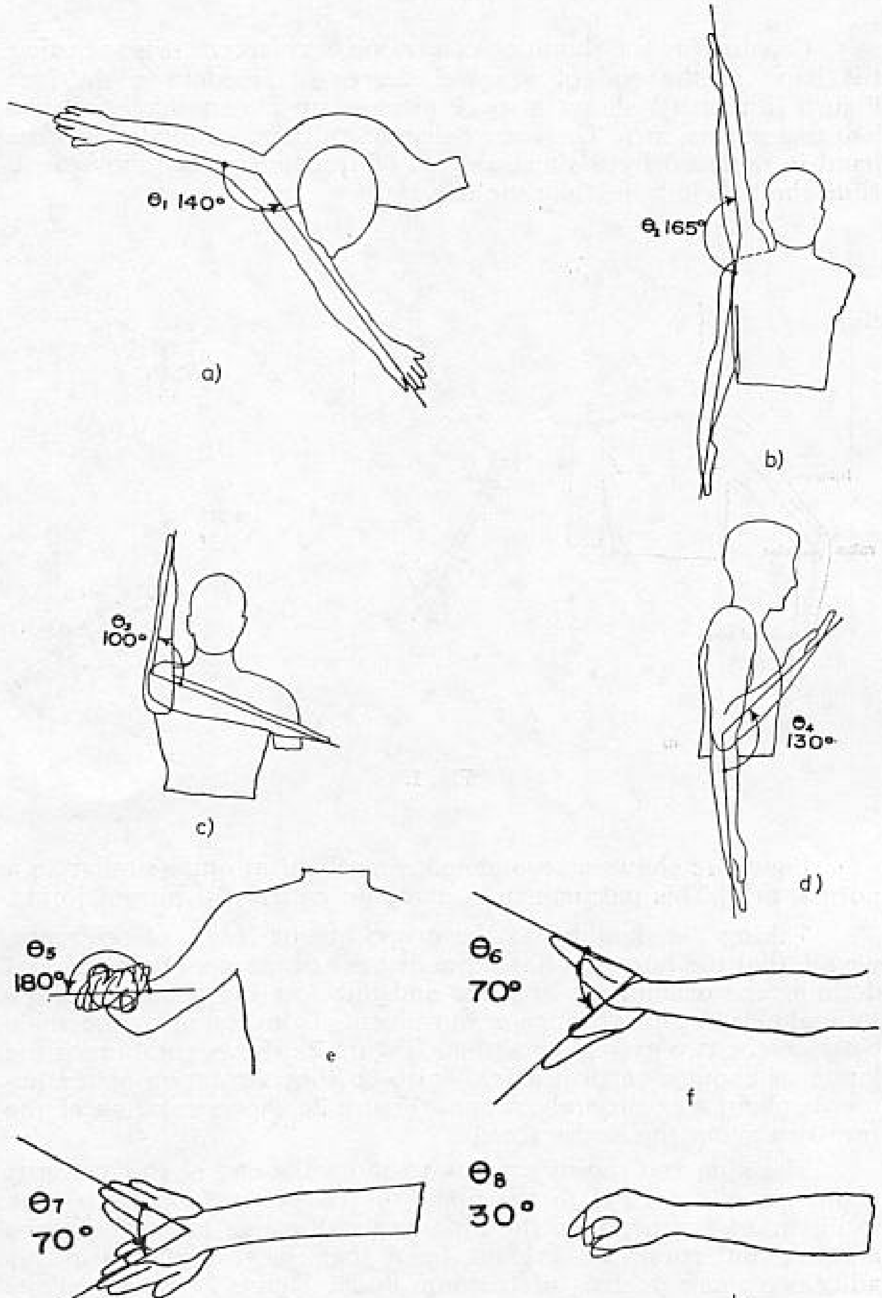


Fig. 2.

Between the hand and the ulna and radius are the carpal bones which provide movement of the hand about two axes, both perpendicular to the longitudinal axis of the ulna and radius. These add two more degrees of freedom θ_6, θ_7, and are shown in Figure 2f and Figure 2g.

Another degree of freedom θ_8 is added by considering the hand as a gripping device such as a hook (See Figure 2h). We thus have eight degrees of freedom in all when we consider the movement of the hand as a single degree of freedom mechanism. In fact the hand considering the fingers only has twenty degrees of freedom.

The forces which can be exerted by the normal arm are quite considerable. It is not essential to equal these forces in an artificial limb. We may set as the limits the force level required in the tasks essential for independence such as eating, dressing, washing, etc. A gripping force of two kilograms and the possibility of raising half a kilogram at arm's length would be adequate for a ten year old child for most purposes.

On the Feasibility of Equalling Normal Arm Performance

The above analysis gives some idea of the complexity of the movements a normal arm can carry out and why a powered hook is such an inadequate device in comparison. When we consider the feedback the hand provides on force, pressure, temperature, texture, slippage, vibration to mention only some of the variables, and that it is self-repairing and usually lasts a life-time without any maintenance it becomes obvious that building a replacement is an extremely difficult task. It is quite safe to say that it is an impossible task for us because it represents the result of milions of years of evolution which we cannot hope to catch up with.

Certain parts of the human body can be replaced with relative ease e.g. teeth. When the part to be replaced performs a sophisticated function with feedback to the brain the task is totally different. It has thus been possible to replace the action of the heart, the kidneys by machines for brief periods, and even in this case of simple functions such as pumping or dialysis the problems are far from solved. It has not been possible to do the same for a hand or leg because of their kinematic, kinetic and feedback characteristics.

The problem facing the designer of replacements for parts of the human body are therefore amongst the most challenging in the field of engineering technology precisely because the design specifications are so demanding.

The Actuator

I considered the piston actuator to be the principal stumbling block in the way to further improvements in limb design. It seemed that a rotary actuator would fit in better with the kinematic requirements because all the articulations in our body consist of rotations about various axes. It did not seem possible to improve significantly the design of existing limbs commercially available unless a completely new approach was taken.

If we consider powering an elbow joint it is seen that as one part is rotated relative to the other a certain volume is swept out which theoretically could be used for the working fluid (See Figure 3).

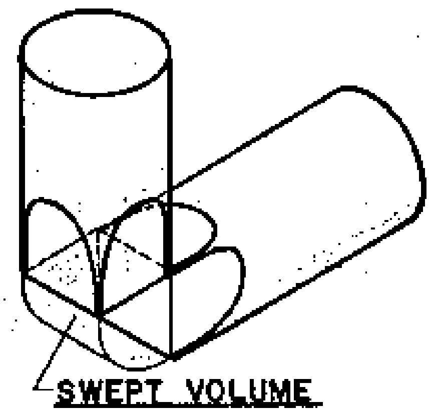


Fig. 3.

A toroidal bellows would provide a solution if means of preventing radial and lateral buckling could be found. Unfortunately the technical difficulties of carrying this out in practice on the small scale required for limbs made this approach unrealistic.

The idea then occurred to me to use a number of inter-connecting inflatable pouches radially restrained at the hinge axis which rotate the restraining blocks relative to one another with the introduction of fluid. Nylon film, polyurethane, PVC and rubber coated fabrics proved suitable material for the pouches.

The construction of a four pouch actuator [9] is shown in Figure 4.

Because the actuator is a new machine element, the design equations are given below.

In this approximate derivation only tensile stresses due to internal pressure are considered.

The moment angle function of the pouch actuator is dependent on supply pressure, axial and radial length, number of pouches and modulus of elasticity of the pouch material. If we neglect the

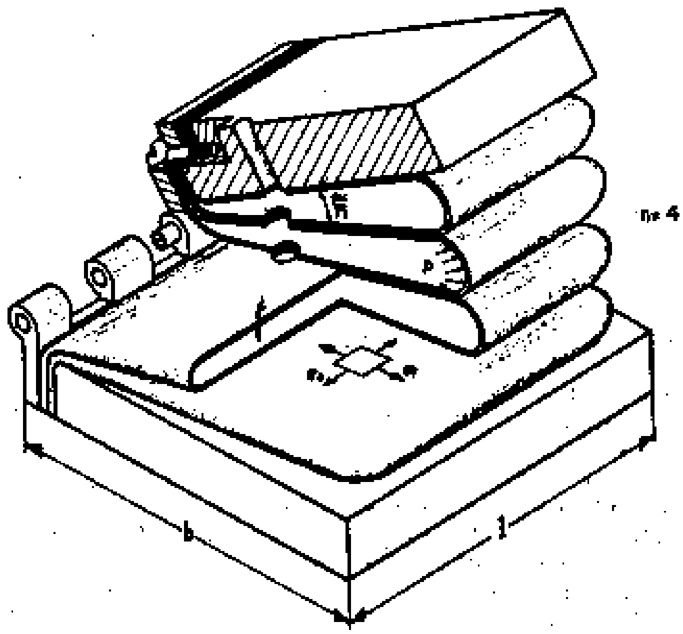


Fig. 4. Pouch actuator

small deformations of the material we obtain the following approximate expression for the moment.

$$M = \frac{1}{2} p b^2 L \left[1 - \frac{\alpha}{n} \left(1{\cdot}6 + 1{\cdot}06 \frac{b}{L} \right) + 1{\cdot}65 \frac{\alpha^2}{n^2} \frac{b}{L} \right]$$

The parameters in this equation and those entering in the design of the actuator will now be defined.

U — energy output required of actuator
α — angle of actuation
M — average moment of actuation
n — number of pouches
t — thickness of pouch material
b — radial length of pouch
L — axial length of pouch
σ — design stress of pouch material
P — supply pressure
Q — volume of pouch material required

Connecting these ten parameters are the following four independent equations (for small values of $\frac{\alpha}{n}$)

$$1.\ M = \frac{U}{\alpha} \qquad 2.\ Q = 2\,b\,t\,L\,n \qquad 3.\ M = \frac{1}{2}\,p\,b^2 L$$

$$4.\ \sigma = \frac{1}{2}\,p\,\frac{\alpha b}{n t}$$

From these four equations we can derive a relation between Q, σ and U:

$$5.\ Q = \frac{2\,U}{\sigma} \qquad U = \frac{\sigma\,Q}{2}$$

This is an extremely interesting equation from which we observe the very simple relation between the quantity of material required for the pouches and the energy output of the actuator. Taking a design stress of 300 kg/cm² for the polymer film used in the pouches we obtain values of 150 kg cm per cm³ of pouch material. Because the backing plates of the actuator form part of the prosthesis structure we can reduce very considerably the weight of the actuators relative to the prosthesis as a whole.

Because there are ten parameters and only four independent equations relating them we may specify six of them to suit our design and determine the others from the equations. If say, U, α, σ, and p are set by design requirements and the material and supply pressure available, we may still set B and L to suit space requirements and then find M, t, n, and Q from the equations. This allows very wide freedom in adapting the actuator to particular needs.

Other features which make this actuator suited to powered limbs are:

1. *The simple shape of the individual pouches.* This makes it possible to produce them by heat-welding thermoplastic polymer films.

2. *The open to closed ratio of the actuator.* This is defined as the length of the arc traversed by the plate extremity to the pouch stack. Ratios of 20 are easily attainable depending on materials and design requirements. By contrast reinforced bellows are limited to ratios of 3 in linear motion. The piston is limited to less than 1 and the McKibben muscle to about 0.2.

3. *Stability.* The construction of the rotary pouch actuator makes it an inherently stable structure within wide limits. By contrast a linear bellows is very unstable and requires a separate guiding or constraining mechanism.

4. *Efficiency, friction and stiction.* These three factors are connected together. This actuator has very low losses because of

the small ratio of hinge pin diameter to radial length of pouch. Efficiencies of over 95 per cent are attained in even the small size. This enables a very smooth movement of the prosthesis to be obtained in contrast to the jerky motion associated with piston actuators.

5. *Tolerances*. The precision required in making the various components for this actuator is less than that needed in those where sealing depends on the accurate fitting of sliding components. This means simpler installation and maintenance.

6. *Stress distribution*. Figure 5 shows the tensile stresses for an individual pouch. The radial stresses σ_r are approximately constant and equal to pab/2 tn. The axial stress σ_a varies from zero at the hinge axis to the value of the radial stress at the open end. There are bending stresses too but these are small when the pouch is open.

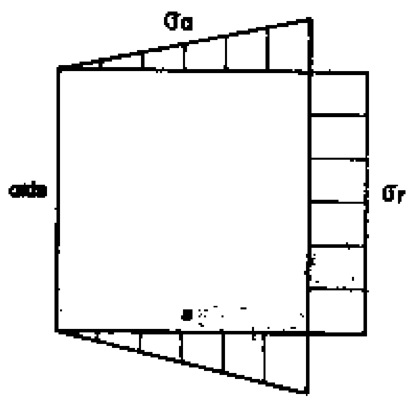


Fig. 5.

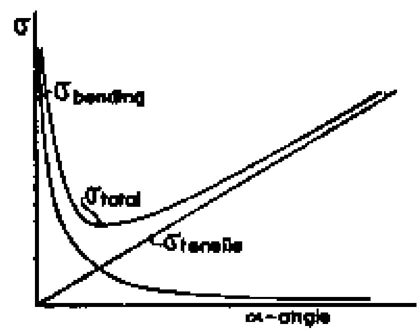


Fig. 6.

On the other hand when the pouch is almost closed, the bending stresses are larger but then the radial and axial tensile stresses are small so that the overall stress is not excessive within the designed operating range (See Figure 6).

The Kinematic Design for a Six Degree of Freedom Arm

The features of the pouch actuator reduced radically the problem of incorporating in the artificial limb the same number of movements in the prosthesis as exists in a normal limb. As a result the main problem becomes the control of the numerous movements rather than the incorporation of enough movements.

I decided to exclude the carpal movements θ_6 and θ_7, (Fig. 1c) as these were not essentail and it seemed that the extra motion did not justify the added control complications.

The degrees of freedom of the prototype limb numbered six and are shown in Figure 7. The orientation of the actuator block was determined rapidly by means of balsa-wood mock up using plastic hinges.

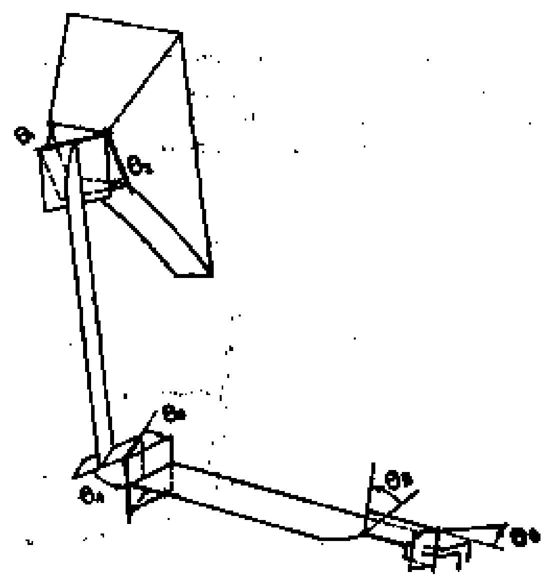


Fig. 7.

It will be noticed that instead of incorporating an actuator with its axis along the humerus to rotate the ulna and radius about it, it was placed at the elbow. When the elbow is at 90° ±45° degrees with respect to the humerus the difference in the kinematics are not significant. The advantage with this arrangement is the much simpler design of the shoulder and elbow unit which consist each of two actuators set at right angles to one another.

Of course it is possible to incorporate in the humerus an actuator to give the correct movement as is done with the wrist rotator to be described below.

At the extremity of the forearm is a wrist rotator and gripping unit. The axis of the wrist rotator (which simulates the rotation of the hand due to the twisting of the ulna and radius) lies along the forearm. One block of the rotator is fixed to the supporting tube.

The other block carries one block of the grip actuator (Figs. 8a and 8b).

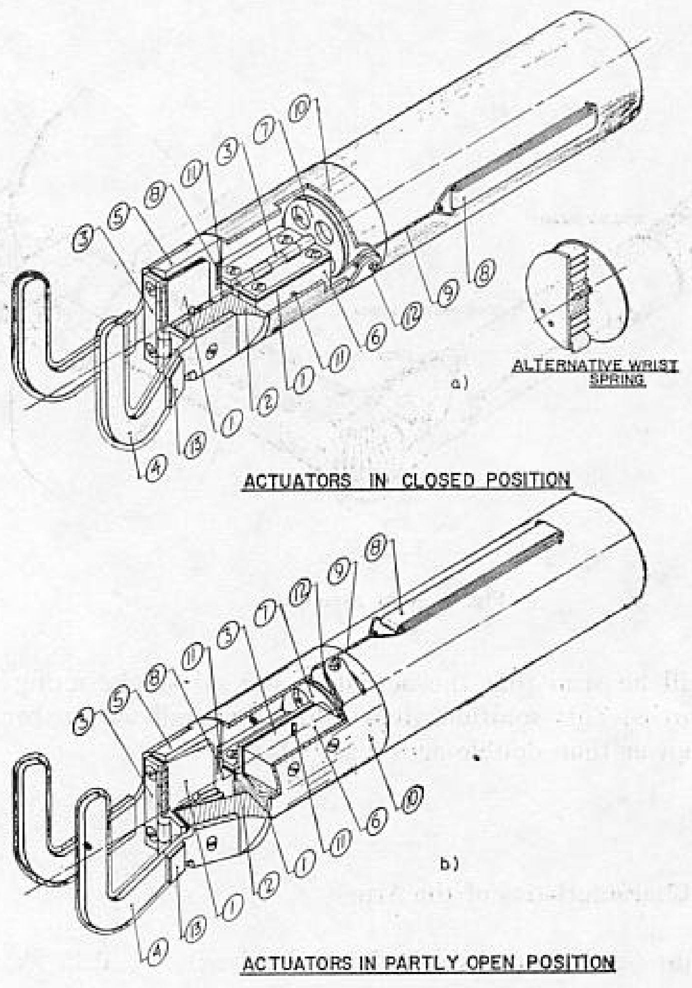


Fig. 8. Wrist rotator and hook assembly
1. pouches; 2. moving hook block; 3. hinge; 4. hook; 5. hook, wrist rotator block; 6. fixed wrist rotator block; 7. return cable pulley; 8. rubber tension spring; 9. cable; 10 wrist cylinder; 11. inlet to actuator; 12. idler pulley. 13. photo-elastic gauge.

The assembly of the shoulder, elbow and hand units are shown in Figure 9. The compressed gas cylinder is shown forming part of the structure of the upper arm.

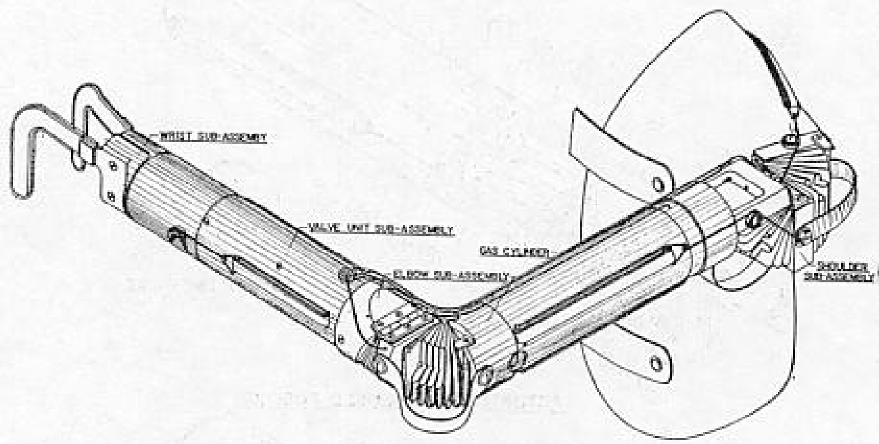


Fig. 9. Arm assembly.

It will be seen that the actuators are all single acting with spring returns. This solution gives a lower overall weight for this range of power than double acting actuators.

Principal Characteristics of the Arm

Weight (excluding gas cylinder and valves)	0.35	kg
Weight of gas cylinder (charged with 100 gm CO_2)	0.375	kg
Weight of six manually operated valves	0.025	kg
Weight of six solenoid operated valves	0.160	kg
Working gas pressure	3.5	kg/cm^2
Hook prehensile force	2.0	kg

Wrist rotator torque	8	kg cm
Elbow actuator torque	45	kg cm
Shoulder actuator torque	65	kg cm
Length of arm (shoulder to hook extremity)	46	cm
Length of forearm (elbow to hook extremity)	28	cm
Maximum net load with arm horizontal	0.70	kg
Maximum net load with humerus vertical and forearm horizontal	0.75	kg

It should be mentioned that by increasing the pressure the operating forces can be increased further. This can be done by increasing the number of pouches in the actuator or by using a stronger material for the actuator. The above data are based on actuators made by heat-welding nylon film.

The Springs

Natural rubber in tension is chosen as the spring material because it gives the maximum energy to weight ratio (about 30 kg cm per gm). This is an important factor in reducing the overall weight of the limb. In contrast steel springs have an energy to weight ratio of the order of 2 kg cm per gm.

In the prototype limb, the wrist rotator has a steel spiral return spring. In spite of the low energy requirements of this unit, its spring weighs more than the combined weight of all the others. In a later version, the steel spring has been replaced by a rubber spring with a saving in weight of about 30 gm (See Figure 8a).

In general the spring force is about one third the actuator force giving a ratio of two to one for the torque in the powered direction to that in the return direction. The spring force is aided by gravity in the case of the shoulder actuator θ_2 and depending on the position of θ_2 also in the case of the actuators θ_3, θ_4 (Fig. 7).

The springs consist of ordinary natural rubber bands in parallel. Bands were found to be most convenient for trials. The bands are anchored at one end along the supporting structure by means of a special hook made of steel wire. At the other end they are slipped into a triangular hook made of soft steel wire and shaped so as to hold the nylon tension cable at one end (Fig. 10). At the other end the cable is fastened to the actuator block moving relative to the structural member to which the spring is fastened. When the angle of actuation is large the cable passes over a pulley to prevent too great a reduction of the effective torque of the cable.

The spring force can be varied by changing the pretension, the number of bands and the effective radius. The dimensions of the rubber bands are (with the exception of the hook spring)

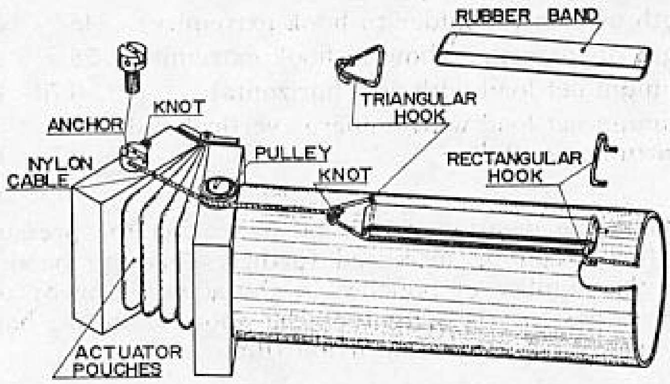


Fig. 10.

Unextended length	50 mm
Width	10 mm
Total thickness	2 mm

Four bands are used for the shoulder actuator, three for the other shoulder and elbow actuators and two for the unit rotator.

The hook uses a short segment of rubber tubing

Internal diameter	6 mm
External diameter	10 mm
Width	8 mm

The rubber bands are pre-extended about 4 cm with a further extension of 3 cm when the actuator is moved. This causes an increasing return force which is partially compensated for by the reduction of the arm of the force as the actuator opens. In the case of the wrist rotator this is not the case and the torque decreases as the angle changes. This is also common to a normal arm (See Figure 11).

The Optical Gauge for the Hook

One of the problems associated with gripping objects by means of a powered hook is the difficulty of applying an appropriate force on the object held. In holding a cigarette or other crushable object the lack of feedback proves a serious hindrance.

Photo-elastic coatings in stress analysis are well known and it occurred to me to use the principle to estimate the stress near the point where the hook is fastened to the rotator block. This allows the user to estimate the force visually by observing the fringe pattern in the gauge.

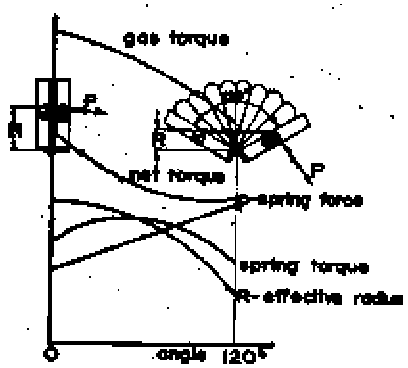


Fig. 11.

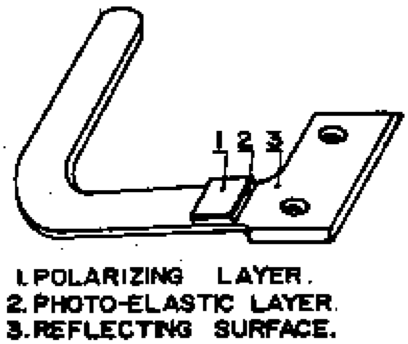


Fig. 12.

The gauge consits of a piece of photoelastic material approximately $10 \times 6 \times 2$ mm in size bonded with epoxy adhesive to the aluminium hook surface of the photoelastic material. When the hook is strained an interference pattern representative of the state of strain may be observed, the polarizing layer acting as polarizer for the incident light and as analyser for the reflected light. Gauges are bonded on both hooks to facilitate observation by the user (See Figure 12).

The Valves

A further advantage of the single-acting actuators is the much simpler valving and piping required. The Kiessling valves were considered heavy and weigh over 150 gm. for a six degree of freedom prosthesis.

A new valve was designed which weighs 25 gm for a battery of six in the manually operated version. The operating force is of the order of 15 gm and the stroke 2 mm which is considerably less than that required for the Kiessling valve. In a solenoid operated version a pair of valves weighs 27 gm of which the two solenoids make up 22 gm. The operating power is 150 mA at 6 V.

The two versions are shown in Figures 13 and 14.

The operation of the valve is based on the deflection of a thin impermeable flexible diaphragm.

The valve consists of one block with two holes in it, a second block with one hole in it and a diaphragm between the two blocks. The two blocks are flat along the plane of contact with the diaphragm and are pressed together by means of screws. The two holes

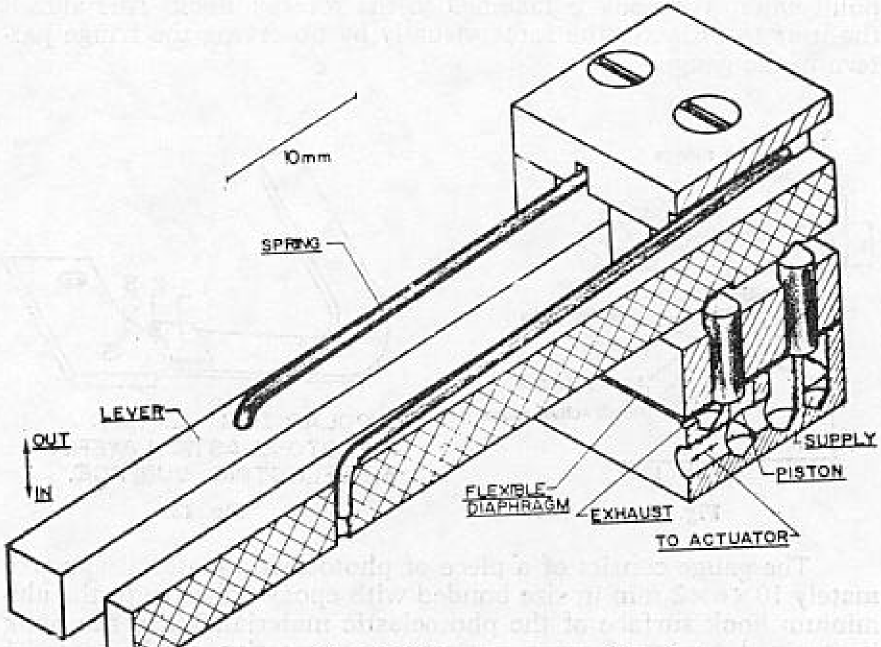


Fig. 13. Two-way pneumatic valve

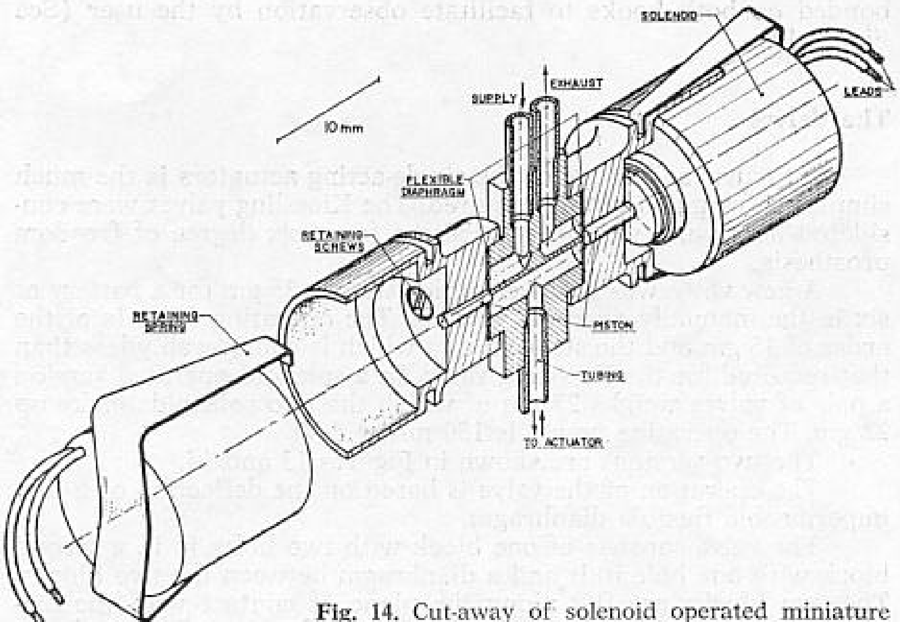


Fig. 14. Cut-away of solenoid operated miniature pneumatic valve.

in the first block are very close to each other so that only a thin web separates between them. These holes are connected to the line whose flow must be controlled. The hole in the second block is symmetrical with respect to the two holes in the first block. In the hole of the second block is placed a cylinder which pressed against the diaphragm creates an uninterrupted plane surface with the second block which effectively seals off the two holes in the first block.

When the cylinder is retracted or the force acting on it removed, the pressure in the line will deflect the flexible diaphragm allowing the gas to flow into the second hole which is connected to the downflow line. Moving the cylinder back will cut off the flow because of the plane surface recreated.

The valve block is made of aluminum alloy and the diaphragm is rubber coated nylon fabric 0.15 mm thick. The cylinder diameter is between 1.5 to 2 mm for the prosthetic application.

For myo-electrically controlled limbs, the solenoid-valve assembly is contained in the hollow tube of the fore-arm as shown in Figure 15.

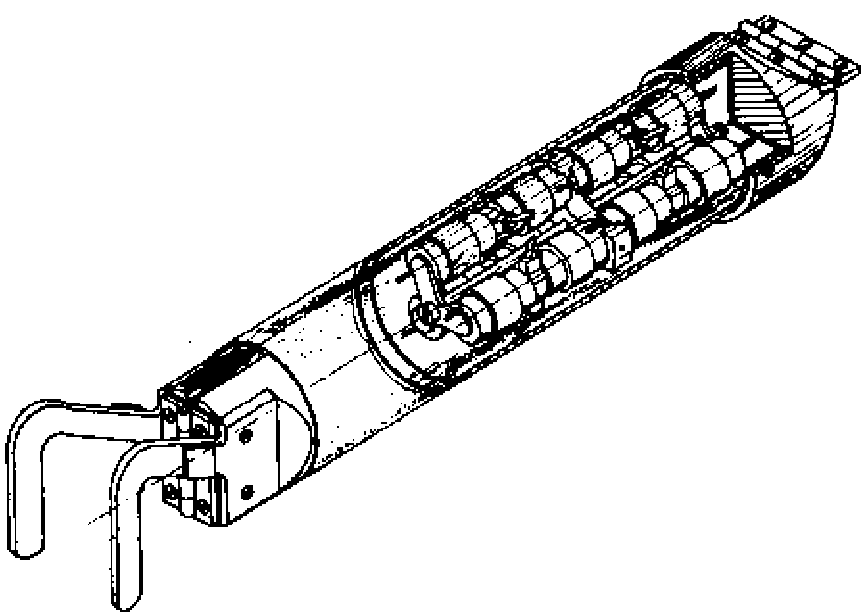


Fig. 15. Solenoid valve assembly.

The Gas Cylinder

The replacement of gas cylinders is one of the factors contributing the most to the cost of maintenance of the limb. It was felt that a desirable solution would be to have the energy source incorporated into the artificial limb and to provide means of recharging the cylinder in situ Specifications were drawn up and with the cooperation of Reynolds Tube Company and Messenger & Sons Ltd. of Birmingham an aluminum alloy cylinder incorporating a filling valve, a reducing valve and a safety valve was designed and built. The cylinder is filled by means of a special probe from a large CO_2 cylinder.

The cylinder is recharged in a manner similar to that used for gas lighters. The probe is inserted into the inlet orifice, the large gas cylinder valve is opened filling it in a matter of seconds, the valve is closed, extra pressure is relieved by means of a relief valve and the probe removed.

The weight of the cylinder is 275 gm and its capacity is 100 gm of CO_2. The maximum weight is therefore 375 gm. Because of the weight of the structural member which it replaces is approximately 70 gm the extra weight excluding gas due to the cylinder being incorporated in the arm is about 200 gm.

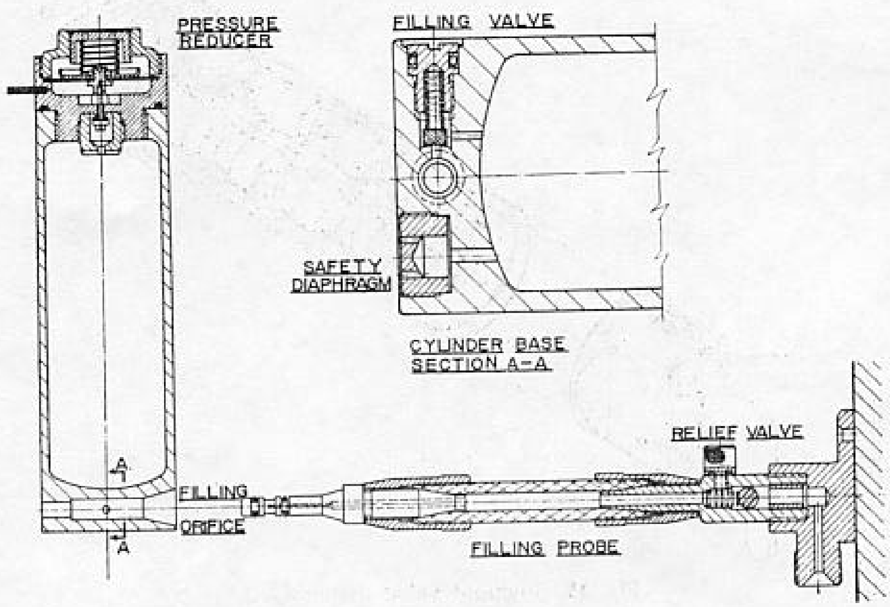


Fig. 16. Gas cylinder

The design of the cylinder and filling probe is shown in Figure 16.

Evaluation

The prototype limb has been tried out on an amelic 8 year old girl. Her lower limbs are grossly deformed and she is usually restricted to a wheel chair. This made it practicable to operate the control valves by means of her toes. One unexpected result of the trial was the way the flexibility of the limb affected her. In contrast to her previous limb which was more or less rigid, the low friction

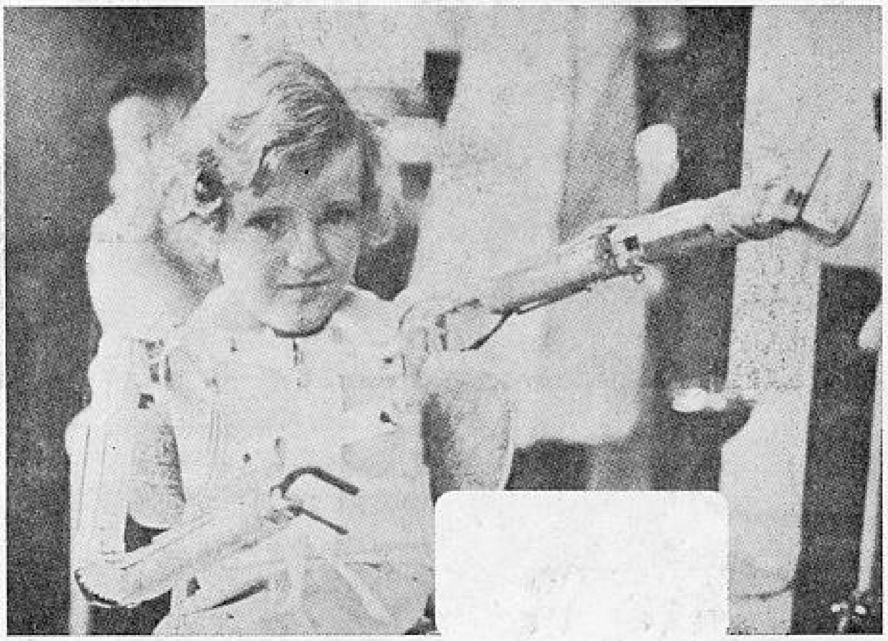


Fig. 17.

of the system and the springiness of the compressed gas in the actuators made it possible for her to move her arms up and down. This she did for a long time apparently enjoying the „feel" of an arm she could wave by moving her shoulders (See Figures 17 and 18). Figures 19 and 20 show the limb in various positions.

A number of arms based on the design described here are being prepared for further more prolonged trials.

Future Objectives

Improving the performance of powered arms depends on control aspects to a great extent.

The use of myo-electric signals is under investigation and provides an elegant solution in certain cases such as amelic children

Fig. 18.

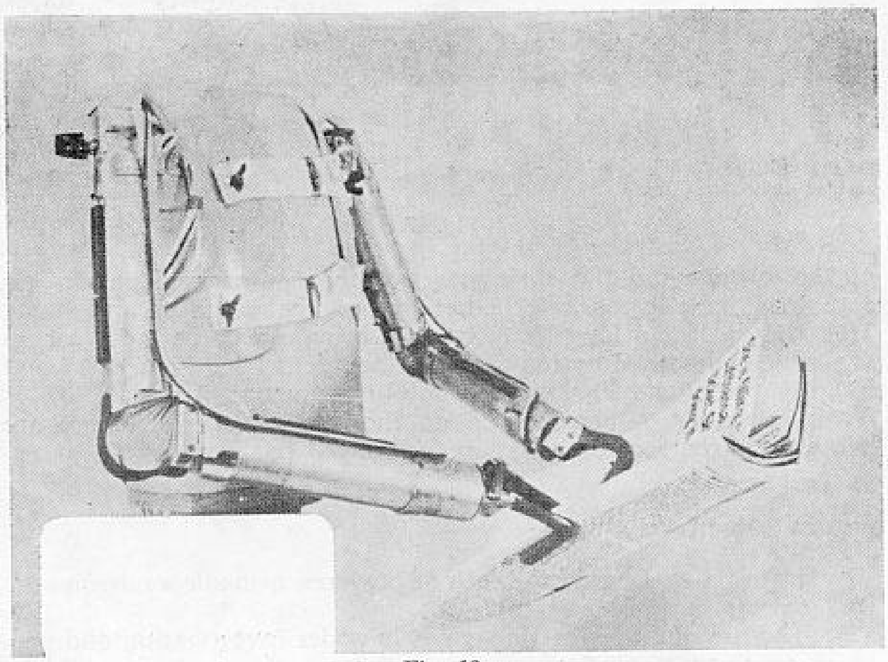


Fig. 19.

and amputees. It seems that a hydrid system consisting of electronic control elements and pneumatic power elements may provide the best overall solution in the near future.

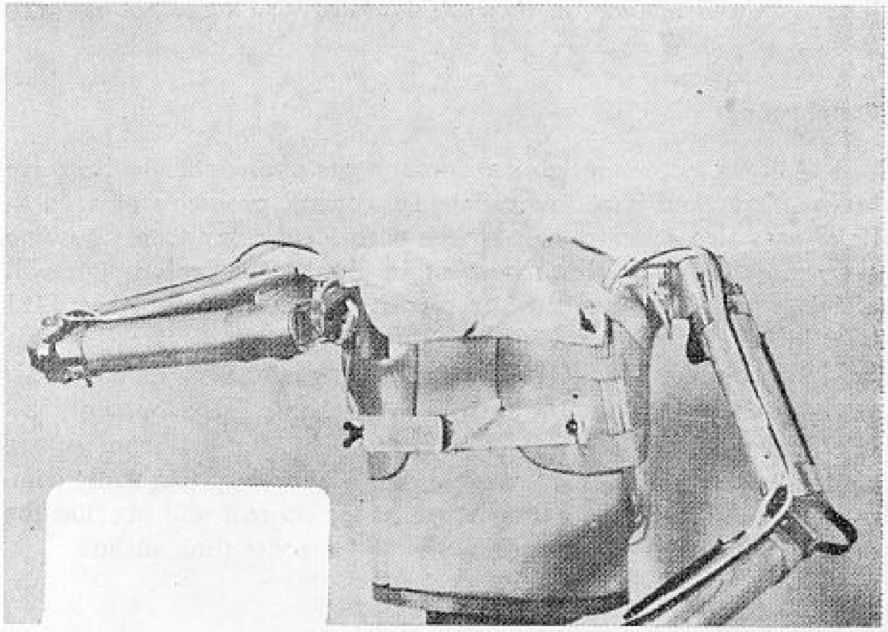


Fig. 20.

A method for simplifying the control of the limb proposed by D. Graupe [10] requires only one command-signal source (mechanical or myoelectric) which serves a reinforcement evaluator for the overall performance of the arm in executing a given task. The evaluation feedback loop is closed through the eye or brain of the user who decides on the reinforcement signal which is of discrete values +2, +1, 0, −1, −2.

A logic-gates device calculates coefficients of regression equations relating by a method of least squares the last change in performance evaluated by the human brain according to the task to be performed with the corresponding changes in the various degrees of freedom during the previous control interval. Reduction of variables may be possible at an a priori programming level by dividing all possible tasks into say 4 major modes each with its own set of control variables since all the degrees of freedom may not be required at all modes. Fine prosthesis adjustment for completing a task may be aided by body trunk movements.

G. Levy [11] has simulated by means of a digital computer two approaches for solving the problem. The first is based on calculating regression coefficients while the second employs artificial neuron elements. Results obtained so far are encouraging enough to carry the investigation through the hardware stage.

Conclusions

As long as the mechanical components of an artificial limb are bulky, heavy and have low efficiency, control problems of an artificial arm already extremely severe with ideal components became even more difficult. Improvement in the mechanical design will alleviate some of the control problems. The recent symposium [12] held in London on problems of artificial limbs indicates that this field of science is still in the pioneering stage. A considerable amount of work remains to be done in mechanical developments and miniaturization before full benefits may be derived from future control systems. It seems probable that a hybrid system using compressed CO_2 for power and electronics for control will provide the best overall solution for artificial limbs for some time ahead.

Acknowledgments

I would like to express my thanks to ICI for the research fellowship during which this project was begun and to Lady Mary Hoare, Brigadier George Chatterton and Mr. Eric Goudie for the important financial support provided by the Lady Hoare Thalidomide Appeal and the Goudie Trust, to Dr P. Mitchell, Dr. P. R. Palmer, Mr. R. Woolgar and Mr. E. Smith of ICI for technical help and advice at various stages of the project, to Dr. P.J.R. Nichols and Mr. T. Strange of Mary Marlborough Lodge, for useful exchange of ideas, to Mr. C.R. Turner and Mr. T.R. Yates for help in designing the gas cylinder, to Mr. D. Stewart for help with the actuator pouches to Mr. F. Beviere, Mr. R. Manton, Mr. Z. Ginot, Mr. P. Schechter and Mr. P. Meisel for their help and suggestions in building the various components and to Mr. R. Davidas for his excellent drawings. Thanks are also due to Mr. J, Waggott and D.R.W. May for their help in fitting the limb and last but not least to little Tina Gallapher for her cooperation in trying it out.

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