An innovative shoulder complex with two active axes for artificial upper limbs

B. Cattaneo, F. Casolo, M. Camposaragna, V. Lorenzi
Sez. S.T.M. Dept. I. S. - Politecnico di Milano, Italy

Introduction

The present total upper limb prostheses, except for an exiguous number of prototypes, have passive shoulder joints, i.e. they need an external action to place the shoulder in the desired position. Weight, size, noise, unnatural driving and controlling procedures of present shoulder prototypes are the main causes that prevent their implementation in the clinical practice. Moreover, a total artificial upper limb system is very complex, since it must also include artificial elbow and hand and must hence sustain a great load and requires a sophisticated driving architecture. For that reason, and also because shoulder amputations are less frequent than finger, hand and elbow ones, researches concerning active shoulder joint prostheses are very few.

Nevertheless we believe that many patients, especially if they are bilaterally disarticulated, would benefit of active, motor driven shoulder joints. Aim of our research is to overcome the above said limitations and problems by means of an innovative approach to the mechanical and electronic design of the system. First of all, it has to be noticed that the present few working prototypes of shoulder joint have only one driven d.o.f. in spite of the physiological three d.o.f.: it helps to keep the system rather simple and lightweight, but permits unnatural and restricted movements. As a satisfactory trade off, we established to develop the artificial shoulder joint with two driven rotational axes, and optionally the third d.o.f. – the arm axial rotation – is created adding a second axis (arm longitudinal axis) at the elbow joint.

The main requirements of our system are to be lightweight, durable, noiseless and aesthetically acceptable, to be able to move the hand covering most of the physiological working area and to generate fairly natural movements for the upper limb; essential requisite of the prosthesis is to be easily driven by the patient, without requiring long training or causing mental stress. As concerned with the loading capability of the arm, it can be reduced with respect to the natural limb, because an amputee will never want to carry an excessive weight due to the fact that high loads on the hand generate relative movements at the body-prosthesis interface and the high stresses at the connection produce pain to the patient. The prosthesis has to be powered by low voltage rechargeable batteries located inside the artificial limb, hence the power required by the system must be reasonably low in order to keep the batteries sufficiently small and lightweight. The power needed by the system is inversely proportional to its mechanical efficiency, which is also correlated to the level of noise generated. Consequently the mechanical efficiency is a key factor of the artificial shoulder design.

Methods

Each additional active axis means one extra motor plus transmission and consequently extra weight, volume, noise, complexity. Thus, as first step, we carried out some kinematics simulations in order to verify which is the minimum number of active axes required to achieve a sufficiently natural limb kinematics. The results have clearly shown that all the solutions with only one revolute pair at the shoulder joint cannot satisfy most of the basic hand working requirements; on the contrary, the solutions obtained with two axes, for some axes orientation, guarantee the execution of the most important tasks such as objects grasping, moving and placing, or eating and drinking.

From this starting point, we designed two different solutions having as major objectives the mechanical efficiency. For this reason, keeping away from adopting long chains of gears, which are inefficient, we used other kind of transmissions, such as multi bar linkages and ball screw linear actuators. The first tentative mechanism (fig.1) has been designed on the basis of the experience acquired by the authors with an artificial elbow recently produced in cooperation with the Centro
Protesi Inail (BO). It includes a d.c. mini-motor plus an eight bar linkage for one axis, and a stepper motor plus a pulley transmission for the other one. This kind of system has a very high efficiency but it is quite complex and has some other disadvantages, such as, for instance, a non physiological position of the flexion and abduction axes of the shoulder and consequently unnatural movements may be generated. Therefore even if the preliminary simulations show that the system can fulfill the main cinematic requirements, with good efficiency factor, we have been forced to move toward a different solution. The chosen solution is based on the use of two d.c. mini-motors, which, at the end of the transmission, act on two moving gears of a differential mechanism (gears 2 and 3 in Fig.2). The third one (gear 1 in Fig.2) is rigidly fixed to the support on the patient’s trunk. In this way, acting on the pulleys fixed to the two moving gears it is possible to obtain composed movement around a and b axes (fig. 2) lying on the same plane. Gleason gears have been chosen for their efficiency and low noise.

The problem of power transmission from the motors to the Gleason gears has been solved using a sophisticated system described below. The worm-gear transmission has been discarded because of its very low efficiency. In the final solution, each motor transmits the motion to a parallel axis on which it is mounted a linear ball-screw which drives a nut that moves the cable acting on the pulley fixed to one moving gear. Another important design constraint is the need to avoid the indirect motion, i.e. to avoid limb motion caused by the load on the hand when the motors are turned off. Excluding, for this aim, to use active mechanisms, because of the amount of extra energy they require, we applied a special kind of correction to the couple of gears a and b (fig. 3). This produces a gearing with a good efficiency with direct power flux and a very bad one with indirect power flux. It is possible to get a very bad efficiency for the indirect motion correcting the gears in order to obtain a pressure angle very near to the friction angle of the used material. Moreover using a full recess correction it is possible to cut, for the indirect motion, the access phase, which is the more efficient. It is also necessary to prevent damages to the system when accidental high external loads are applied to the limb, as may happen during a patient’s fall. To achieve this goal, some gears of the system are connected to their axis by means of a conic friction (wheel b and axis c – fig.3). In order to obtain an adequate working space for the arm and fairly natural movements with a shoulder lacking one d.o.f. with respect to the natural joint, it is essential to optimize the orientation, with respect to the trunk, of the motionless conic gear’s a axis, and of the elbow axis with respect to the arm.

In our case the position of these axes has been obtained by a multifunction optimization technique. The optimization algorithm is composed by two steps: the first one selects the range of possible axis orientations in order to obtain a working space including patient’s mouth and to avoid mechanism Gimbal Lock. The second step include an optimization function which minimizes the distance between artificial and natural elbow trajectories while hand follows measured paths. Reference movements of some normal subjects have been collected by means of a specialized video-electronic system.

For some movements (e.g. drinking and eating with a spoon) another important condition has been imposed in the optimization process: the capability of the mechanism to keep the hand to a rather constant orientation during glass and spoon displacement tasks.

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**Fig. 2:** Differential mechanism

**Fig. 3:** Shoulder complex
Results & Discussion

The total efficiency factor ($\eta_{\text{tot}}$) of the transmission can be evaluated by the product of the efficiency factors of all the mechanism in series (fig. 3). From the motor axes to the shoulder ones, the efficiency factors are: $\eta_1$ for the epicyclical reducer mounted on the motor (M), $\eta_2$ for the elicoidal corrected pair (a-b), $\eta_3$ for the ball-screw system (c-d) and $\eta_4$ for the cable and pulley system (e), $\eta_5$ for the Gleason conical couple of the differential mechanism (f-g).

The total mechanical efficiency factor, for direct motion, is $\eta_{\text{tot direct}} = \eta_1 \cdot \eta_2 \cdot \eta_3 \cdot \eta_4 \cdot \eta_5 = 0.58$, quite good with respect to the one of other commercial systems (e.g. in a commercial elbow prosthesis we measured an efficiency factor of 0.06).

Moreover, the total efficiency factor, in indirect motion, is $\eta_{\text{tot indirect}} \approx 0.15$. This value together with the braking moment, given by d.c. motors short circuited, and their friction moment, allows to maintain the hand stopped under a 10 N load.

The size of the mechanisms of the described new shoulder is sufficiently small to be easily fitted into the volume corresponding to a normal upper limb. The weight of the full limb included batteries is not higher than the weight of a natural limb.

The optimization process gives an axis orientation that generate sufficiently natural movements keeping the hand at a rather constant orientation for drinking or eating.

The hand workspace representation (fig.4) shows that, with the chosen shoulder axes, the hand can easily reach most of the volume normally used for grasping objects and the volume near the subject’s mouth.

A 5 active d.o.f. total upper limb prosthesis like the one assembled with: the new shoulder, an active elbow and a hand with active wrist and pinch, need an advanced driving system. It cannot be moved by the patient acting directly and simultaneously on the joints through myoelectric signals – as it happens for the present 1 or 2 d.o.f. systems – because it would be too difficult and stressing for the patient; alternatively the joints could be moved sequentially producing slow not aesthetical movements. Thus we are now working to solve the problem in two steps: the first is to set up a method for communicating to the system where the patient will place the hand and a second one to control the active axes in order to obtain the required motion of the end-effector.

The first step is certainly the most complex: for the preliminary tests we use little head and scapular movements to drive the hand –only the grasping function is acted by a myoelectric signal. A virtual reality system, accepting actual head and scapular movement, has been developed: its aim is to optimize the driving algorithm, and to train the subjects.

References


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