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AN ABOVE-ELBOW PROSTHESIS WITH OPTIMIZED COORDINATION

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Abstract

One way to generate coordinated prosthesis motion is to kinematically couple rotations about two or more joints; e.g. elbow and wrist. The motion of the terminal device in space is then determined by the kinematic parameters of the particular mechanism. The resultant fixed pattern can be useful for certain frequent tasks such as eating and drinking.

In order to study wrist/elbow coupling, a computer model was developed which simulates the kinematics of an above-elbow prosthesis with coupled wrist and elbow rotations about arbitrary skewed axes. A numerical optimization procedure then searches for axis orientations and coupling ratio which will generate the closest possible approximation to a prescribed terminal device path.

A body-powered above-elbow prosthesis was constructed to test the results of the computer optimization. It couples unequal wrist and elbow motions about skewed axes, yielding a spatial motion pattern in which the terminal device remains level enough to prevent spilling while simultaneously sweeping closer to the body as the elbow flexes. It appears that this pattern may be more useful for eating and drinking than planar parallel motion.

Introduction

Coordinated control of several joints is one desirable objective in the design of upper-limb prostheses. Perhaps the simplest approach toward this goal is "kinematic coupling", in which mechanisms couple two or more joints in fixed patterns. This method has the advantage of simplicity and reliability, but it is also the least versatile. However, if a pattern can be generated which is useful for several common tasks, kinematic coupling represents an advantage over conventional prostheses in which, from a practical standpoint, only one degree-of-freedom can be actuated at a time.

Kinematic coupling dates back to at least 1860 (1). Since then, this principle has been implemented in many ingenious ways. One of the most common forms has been a parallel pattern, in which the terminal device (TD) is constrained to remain parallel to its initial position throughout the range of arm motion. This capability for level lifting is potentially useful for eating and drinking, and also for maintaining an optimal grasp angle with respect to a horizontal surface.

Parallel kinematic coupling is embodied in all the versions of the Edinburgh children's prostheses (2, 3) and was one of several modes in the Multi-Mode Arm, an above-elbow prosthesis described by the author in an earlier paper (4). In the latter prosthesis, the parallel mode was judged to be the most useful pattern, but it has one drawback: at full elbow flexion, the TD points straight ahead, making eating or drinking awkward. Passively flexing the TD toward the midline of the body is only a partial solution since the TD is then at a poor angle for grasping an object at full extension. A pattern in which the TD remains level while simultaneously flexing toward the midline of the body as the elbow flexes could be a better solution.

This paper describes a computer model which was used to study the kinematics of wrist/elbow coupling in depth. The results were applied to the design of an above-elbow prosthesis which may offer an improvement over planar parallel motion.

Kinematic Model

Kinematic studies were initiated in an attempt to find a coordinated pattern which is more functional than planar parallel motion. A model of the upper limb was programmed in a digital computer to allow a variety of above-elbow prostheses to be simulated. As shown in Fig. 1, both the wrist and elbow axes of the simulated prosthesis may be oriented arbitrarily in space by specifying the direction cosines of the axes ($\underline{u}_1$, $\underline{u}_2$). Rotations about these axes (ϕ_1, ϕ_2) may be specified, as well as motion of the residual limb in flexion/extension and ad/abduction. Limb segments, whose lengths were derived from anthropometric tables for a 50th percentile male, are modeled as lines and the TD as a right triangular prism to aid visual interpretation of graphical output.

Output from the model is in two forms. Three orthogonal views of the schematic model portray the incremental motion of the prosthesis as the elbow flexes for qualitative interpretation, and a plot of the change in the three Euler angles of the TD vs. elbow flexion furnishes quantitative information. As defined by Goldstein (5), the two Euler angles ϕ and θ measure the relationship of the TD to the horizontal (x-y) plane and ψ measures the rotation about a vertical (z) axis. If we call changes in ϕ or θ "tilt" and a change in ψ "sweep", the objective is to minimize tilt and maximize sweep, i.e.

$$\begin{array}{l} \text{minimize } \Delta\theta,\ \Delta\phi \\ \text{maximize } \Delta\psi \end{array} \qquad (1)$$

The underlying rationale is that a small amount of tilt, perhaps up to 20°, is acceptable for practical purposes, if accompanied by sweep. Allowing the wrist and elbow axes to be skewed relative to each other is one way to achieve this goal.

Optimization Procedure

Optimization was a two-step process. First, the orientation of the elbow axis was studied. Virtually all prosthetic elbows have their axes orthogonal to the humeral axis, so that the forearm moves in the plane defined by it and the upper arm. The human elbow axis, however, is at approximately 80° to the humerus, the "carrying angle" (6). Simulations of various prosthetic carrying angles led to an 80° elbow axis. As a result, the forearm lines up with the upper arm at full extension but angles 10° toward the midline of the body at full flexion. This is intended to position the TD in a more functional location.

Having defined the elbow axis, wrist orientation was studied to optimize TD motion. For simplicity and practicality, ϕ_1 (wrist flexion) was constrained to be linearly proportional to ϕ_2 (elbow flexion); i.e.

$$\phi_1 = \kappa\phi_2 \qquad (2)$$

where: κ is a constant coupling factor

Therefore, the unknowns in the optimization process were κ and $\underline{u_1}$, the orientation of the wrist axis, which is:

$$\underline{u_1} = \begin{bmatrix} u_x \\ u_y \\ u_z \end{bmatrix} \quad (3)$$

where: u_x = cosine of the angle between the wrist axis and the x-axis, etc.

Since $u_z = (1-u_x{}^2-u_y{}^2)^{\frac{1}{2}}$, this represents minimization of a function of three unknowns. By formulating an objective function in terms of tilt and sweep, a "Flexible Simplex" (7) numerical optimization procedure could be used to search for values of the unknowns which would minimize that objective function.

Results

The solution that the computer converges upon is dependent upon weighting factors in the objective function. Therefore, many possible solutions were generated, each one an optimum for the particular combination of weighting factors. As might be expected, a trade-off between tilt and maximum sweep exists: solutions with greater sweep also have greater tilts. However, solutions with acceptable tilt (less than 20°) and substantial sweep (almost 80°) could be found.

Examination of the various solutions showed that the motion of the TD is very dependent on the supination angle of the wrist (θ_x), while the angle between the wrist axis and the long axis of the forearm (θ_z) changes little from one solution to another. This suggested the possibility of a prosthesis with a wrist axis which could be passively pronated or supinated to alter the motion characteristics of the TD. It was subsequently discovered that θ_z could be fixed at 80° and θ_x allowed to be adjustable from 90° (wrist and elbow axes parallel) to 50° (supinated 40°). For this configuration, the optimum coupling ratio is 1.08; i.e. $\phi_1 = 1.08\ \phi_2$.

The result is a prosthesis with variable sweep, as shown in Fig. 2. With zero supination, a parallel pattern is approximated over the 140° range of elbow flexion, with no sweep and 11° maximum tilt (due to a non-unity coupling ratio). Fig. 3 shows the computer output for this configuration; the effect of the "carrying angle" is apparent. At 40° supination (Fig. 4), the TD sweeps 77° with a maximum tilt of 19°. Intermediate supination angles yield intermediate values of sweep and tilt, according to the graph in Fig. 2.

All of the simulations described above assumed no motion of the residual limb. Simulations incorporating typical values of humeral flexion showed that it actually tends to decrease the amount of tilt. Therefore, the actual system on an amputee should generate an even closer approximation to level lifting.

Prototype Prosthesis

In order to test this concept as simply as possible, an endoskeletal cable-operated prosthesis was designed according to the computer-generated specifications (Fig. 5). Pulleys couple wrist and elbow motion with a cable passing through the forearm pylon, and the wrist can be passively rotated to generate varying amounts of sweep as described earlier. An elbow lock is provided so that an amputee can control the limb with conventional dual-control.

As shown in Fig. 6, level lifting accompanied by sweep is realized on a test stand. Tests of the prosthesis on an amputee will determine if these principles actually represent a functional improvement.

Conclusions

Computer simulation of the kinematics of wrist/elbow coupling was undertaken in order to design an above-elbow prosthesis with optimum coordination. The result is a body-powered prototype prosthesis which can lift objects without spilling while simultaneously sweeping closer to the midline of the body. If verified by amputee evaluation, this could represent an improvement in function over fixed-wrist and parallel-pattern prostheses, especially for children and high-level amputees.

Acknowledgements

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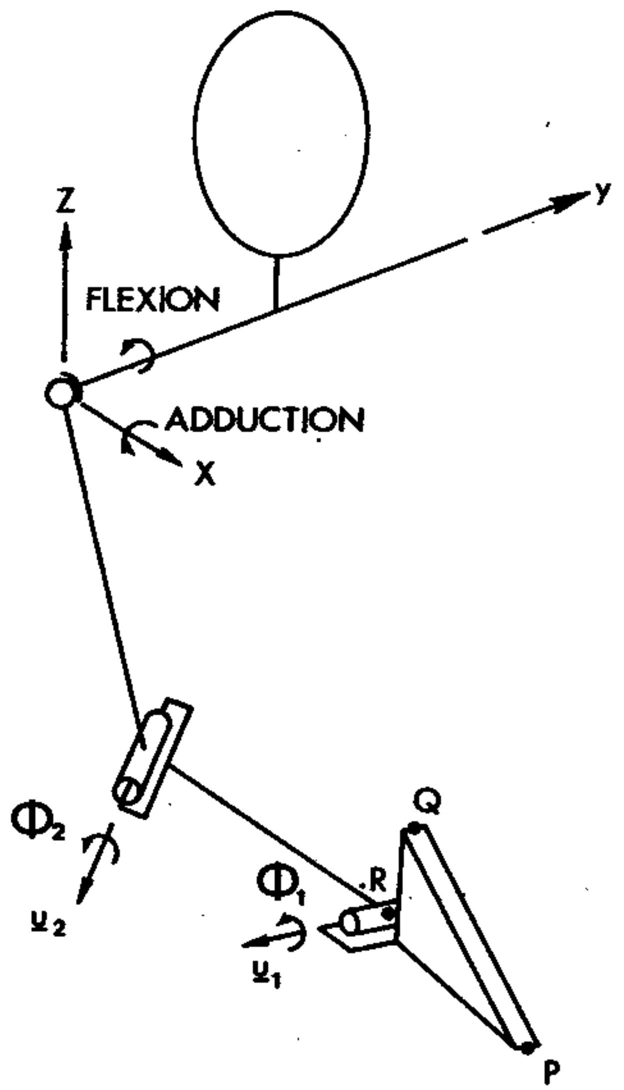

Fig. 1. Kinematic model simulated by the computer.

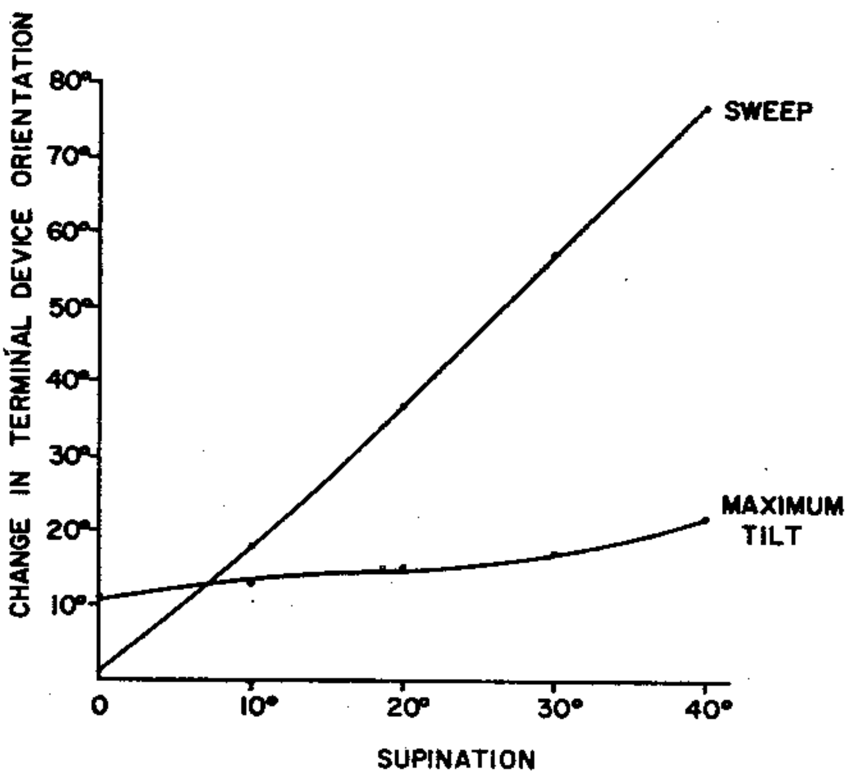

Fig. 2. Changes in terminal device orientation during 140° elbow flexion as a function of wrist supination angle. Zero supination approximates a parallel path; increases in supination cause the terminal device to sweep closer to the midline of the body.

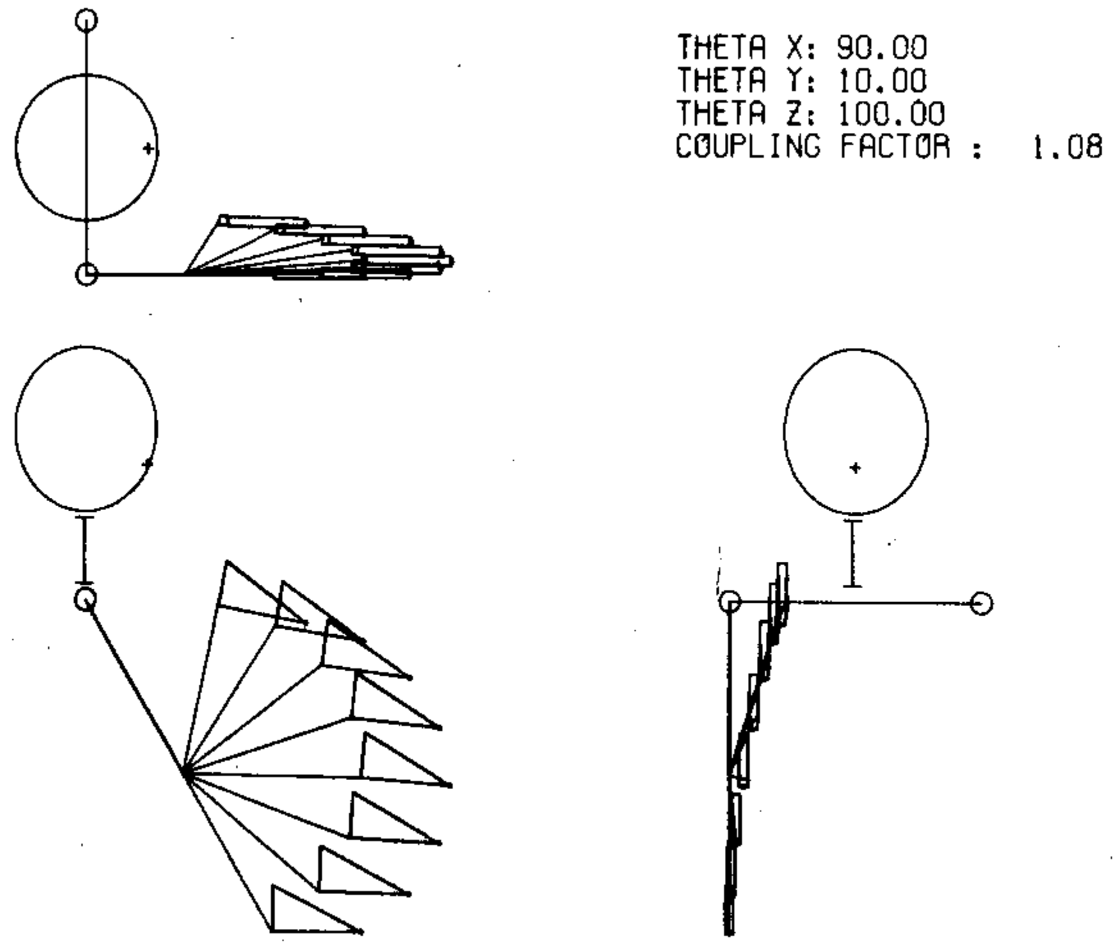

Fig. 3. Motion of the prosthesis with zero supination.

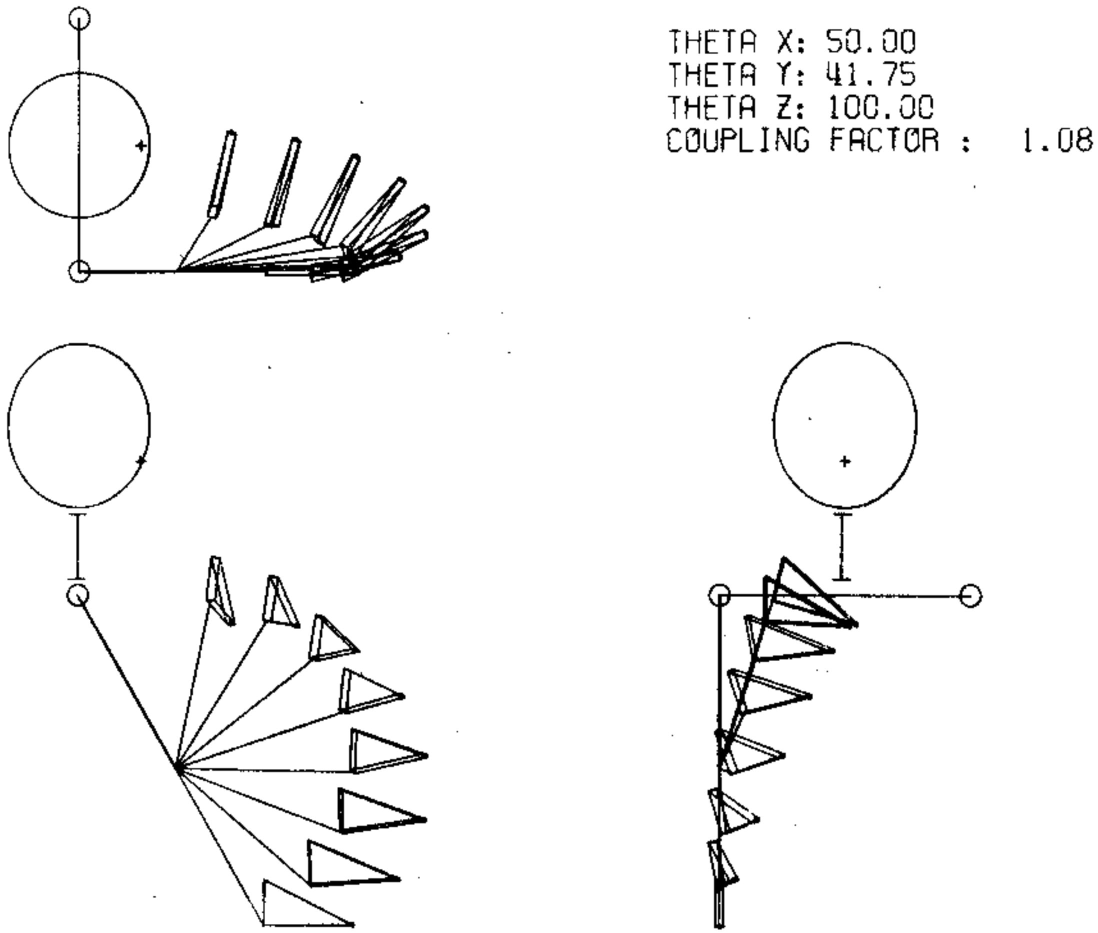

Fig. 4. Motion of the prosthesis with maximum supination (40°).