An Admittance Control Scheme for a Robotic Upper-Limb Stroke Rehabilitation System

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Abstract: This paper presents a control scheme for a dual robot upper-limb stroke rehabilitation system. A model of the human arm is outlined and used to formulate an admittance controller operating in human upper-limb joint space. Initial results are provided together with a discussion of future work.

I. Introduction

A number of robots have been previously developed to investigate and implement stroke upper-limb rehabilitation. The MIT Manus project [1] consists of a SCARA manipulator connected to the patients arm by an orthosis at the wrist. It can provide assistance in two translational degrees of freedom (d.o.f.) in the horizontal plane. A cartesian impedance control scheme modulates movement in this plane.

The Gentle/s project uses a commercially available haptic robot providing three active translational d.o.f. [2]. The robot is attached to the patients forearm using a specially designed orthosis, allowing three rotational d.o.f. at the wrist. A second orthosis, located at the elbow, is suspended from an overhead frame to support the proximal part of the arm. The robot is admittance controlled in each of the active d.o.f.

The MIME series of robots provide bilateral therapy for the upper limb where the patient's less affected arm is used to define the movement of the affected arm [3]. The most recent version, named ARCMIME, consists of two forearm orthoses moving on linear slides, the first is powered and the second acts as a measurement device. The powered orthosis provides variable assistance to the patient using a PID control scheme.

Other robotic systems, such as the ARM Guide [4], use similar approaches to those above. In general, all the systems feature a single point of attachment between the patients arm and the robotic device. The controllers focus on the movement of the attachment point in robot task-space. Varying levels of assistance can be provided to help the patient follow the desired trajectory. However, there is no provision to monitor or control the coordination and orientation of the patients arm.

Learning any motor skill involves achieving appropriate coordination between the movements at the various degrees of freedom of the involved joints, thereby co-ordinating the velocity and position trajectories of the related body segments. For example, forward flexion of the shoulder and extension of the elbow are involved in advancing the hand when reaching [5].

The robotic system under development at Leeds has two attachment points on the patients upper limb, as illustrated in Fig. 1. The orthoses have three degrees of active translational freedom. The orientation of each orthosis is unconstrained, the related rotation axes pass through the arms centre. The result is a total of six powered d.o.f. This gives the potential for alternative control strategies that focus on a coordinated movement of the arm, rather than just the movement of the hand. The proposed robot control system will adapt its assistance on each attempt the patient makes at the desired movement, based upon analysis of the previous attempt. This analysis will form the basis of the control algorithms which will direct the robotic assistance. Therefore it is necessary to specify the arm exercise trajectory in terms of human joint space, rather than robot task space. This requires a kinematic model of the human arm. From this model forward and inverse kinematic calculations need to be resolved to implement position control.

Fig. 1. Orthosis attachment points for the Leeds Physiotherapy Robot. Coordinate frames demonstrate the powered d.o.f. provided by the robotic device.

Impedance and admittance control schemes have been used with success to provide variable assistance to the patient in the robot systems discussed above. They have all operated in robot task-space. A logical extension to specifying trajectory in human joint space is to implement admittance/impedance control in this space. This allows varying levels of assistance to be applied to each d.o.f. of the arm. Our robotic system uses pneumatic actuation, favouring admittance, rather than impedance, control [6]. Admittance control modulates the position trajectory relative to a function of the force measured, (1).

\[
\delta x = f(F_{\text{actual}} - F_{\text{desired}})
\]
Consequently, to implement admittance control in human joint space a measure of the force or torque about each d.o.f. is required. Each of the orthoses seen in Fig. 1. incorporates a 3 d.o.f. force transducer. These force measurements must thus be resolved to human joint space according to the kinematic model.

II. HUMAN ARM KINEMATIC MODEL

A. APPROACH

The kinematic human arm model forms the basis of the controller. A compromise is needed between model accuracy and the ability to find a closed-form inverse kinematics solution. A human has layers of skin, fat, connective tissue, muscle and bone that move dynamically in relation to one another as well as complex joints with variable centres of rotation that result in a highly complex system to model. However, a number of assumptions can be made in order to model the human upper-limb as a set of rigid bodies [7].

- The mechanical behaviour of the upper limb with respect to the trunk is independent of the lower half of the body whilst the person is in the sitting position.
- The deformation of soft tissues does not significantly affect the mechanical properties of the limb segment as a whole.
- Within each segment (forearm and upper arm), bones and connective tissues have similar rigid body motions.

These assumptions allow the use of standard kinematic modelling techniques, specifically the Denavit-Hartenburg notation system widely used in robotics. This provides a convenient method by which coordinate frames can be assigned and the forward kinematics defined.

B. FORWARD KINEMATICS

In the human arm the shoulder complex poses the greatest challenge to model. It is composed of a number of joints which allow both rotational and translational movement of the shoulder girdle[8]. Existing models of the shoulder range from an assumption that it is a simple spherical joint with three degrees of freedom [9], to one of seven d.o.f. [7]. A five d.o.f. model has been selected [10] as a compromise that approximates the compound movements of the shoulder without requiring detailed modelling of the internal elements. This model assumes that the shoulder is a spherical joint gliding on a plane surface, Fig. 2. The spherical joint models the glenohumeral joint whilst the two d.o.f. in translation represents scapular movement in relation to the thorax.

The complete arm model, described in terms of Denavit-Hartenburg coordinate frames, is shown in Fig. 2. Elbow extension and flexion has been modelled as a single d.o.f. hinge joint. The distal orthosis allows freedom of movement about the sagittal axis of the forearm (allowing for pronation and supination), as described in section I.

C. INVERSE KINEMATICS

It is desirable to find a closed form solution to the inverse kinematics problem. This allows the calculations to be incorporated into a high speed control loop without the computational demands typically associated with numerical estimation.

Through inspection it was found that a closed form solution can be found given the following information:

A) The current position of the upper and forearm orthoses
B) The position of the glenohumeral centre of rotation, i.e. the origin of frames 2-3 in Fig. 2, when the arm is in the anatomic position.
C) The distance from frame 4 to frame 5b in Fig. 2.

Ideally, only the parameters in 'A' should vary during movement with the robotic device. These can be measured using the joint position transducers of each robot and it's forward kinematics calculations. The parameters in 'B' and 'C' can be determined by using the robotic device as an unpowered measurement device in conjunction with numerical estimation techniques. Parameter C will remain constant provided the orthosis does not slip on the patient's arm. The arm support currently being developed will ensure that this does not happen. Parameter B will also remain constant if the patient's torso is fixed. The use of a harness is being investigated to limit this movement during robotic physiotherapy.

III. RESOLVING JOINT TORQUES AND FORCES

The final component of the arm controller is resolution of the various human joint forces/torques from the forces measured at the orthoses points, as shown in Fig. 4.

To perform a dynamic analysis, mass and inertia properties for the human arm are required. Although typically these are much more difficult to obtain than the kinematic measurements discussed in section II, since the velocities and accelerations that the paretic arm is subjected to are low during typical physiotherapy treatments it was deemed that the contribution of dynamic effects would be negligible. This is demonstrated in section V. Consequently, a static approximation can be made that requires only kinematic parameters.

The sum of the upper and lower components means to statically resolve forces and torques between task-space and joint-space. It can be partitioned into two components, the first describing linear position, the second describing orientation.

\[
\begin{bmatrix}
    F, \tau_{\text{joint}}
\end{bmatrix}
= 
\begin{bmatrix}
    J(q)_{p, \text{upper}}
    J(q)_{p, \text{lower}}
\end{bmatrix}
\begin{bmatrix}
    F_{\text{task}}
    \tau_{\text{task}}
\end{bmatrix}
\]  

(2)

The torques at the orthosis attachment points will be zero because only the linear position, and not orientation, is constrained. Therefore, only the linear position component of the Jacobian is required. Referring to Fig. 4, the joint space forces and torques are a function of the forces measured at both the upper and lower orthosis attachment points. A linear position Jacobian can be derived for each attachment point from the forward kinematics discussed in section II. Since the Jacobian defines a linear mapping between joint and task space the resultant joint torques and forces are equal to the sum of the upper and lower components.

\[
\begin{bmatrix}
    f_d l
    f_d 2
    \tau_{\theta, 4}
    \tau_{\theta, 5}
    \tau_{\theta, 6}
\end{bmatrix}
= 
\begin{bmatrix}
    J(q)_{p, \text{upper}}
    J(q)_{p, \text{lower}}
\end{bmatrix}
\begin{bmatrix}
    F_x, U
    F_y, U
    F_z, U
    F_x, L
    F_y, L
    F_z, L
\end{bmatrix}
\]  

(3)

The final human joint impedance controller is constructed from the elements discussed in sections II and III. A schematic of the controller is shown in Fig. 3. This highlights the transformation of measurements from robot joint-space to robot task-space and finally to human arm joint-space. The impedance controller is currently of a linear/torsional spring-damper form depending on the d.o.f. it is acting upon.

The most significant facet of this controller is the ability to specify the assistance received by each individual degree of freedom on the arm. For instance, glenohumeral subluxation is a particular risk after stroke. During physiotherapy the integrity of the shoulder complex must be preserved by supporting the upper arm carefully. This can be replicated by specifying high levels of assistance to the shoulders vertical translation 'd{sub}i' (See fig X). This is not possible with a conventional cartesian impedance controller because the arm's d.o.f. are coupled with respect to robot task-space. Increasing assistance in the vertical axis will also impact upon shoulder extension/flexion.

Initial testing of the controller has been undertaken using a multi-body dynamic simulation of the human arm. This uses the same kinematic model as seen in Fig. 2. Segment masses and dimensions are defined using anthropometric data for an adult male [12].

As discussed in section III, the static approximation used to resolve joint forces and torques is based upon the premise that dynamic effects are negligible during typical arm movement occurring during therapeutic exercises. A motion capture system was used to record a typical physiotherapy intervention aimed at facilitating a reach-retrieve motion followed by a hand-to-mouth motion. The trajectory was then transformed into arm model parameters. The arm was then moved through this trajectory. All joints were defined as frictionless and gravitational effects were disabled. The resultant human arm forces and torques represent the dynamic components of the movement. These results were used to compare a full simulation of static and dynamic components with a static approximation.

It is evident that the inertial forces and torques are sufficiently small to be considered negligible. In this case the movement was performed by an able bodied person.
The controller presented in this paper has been designed specifically to allow the robotic system to assist the patient to undertake arm movement exercises specified by the physiotherapist. Initial results in simulation have been promising, illustrating the ability to define trajectory and assistance directly at human joint movement level. However, because the implementation involves human contact the need for controller robustness and stability is particularly important. Both the inverse kinematics and transpose jacobian calculations will be subject to singularities at particular arm configurations. Further analysis will reveal if these lie within the typical operating range of the human arm, and if so what measures can be taken to avoid them. Similarly, the controller will be tested for robustness to error propagation through the kinematic chain and undesired force input. The latter concerns artifacts resulting from the patients involuntary movements, such as spasm, which may occur during the assisted arm exercise.

Future investigation will also encompass an evaluation of the performance of the human-arm joint impedance controller in comparison to other methods. In particular, the ability to apply meaningful assistance and simultaneously controlling the coordinate of movement between joints will be addressed.

**REFERENCES**