

MODELLING HUMAN DYNAMICS IN-SITU FOR REHABILITATION AND THERAPY ROBOTS

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Abstract

This paper outlines some rehabilitation applications of manipulators and identifies that new approaches demand that the robot make an intimate contact with the user. Design of new generations of manipulators with programmable compliance along with higher level controllers that can set the compliance appropriately for the task, are both feasible propositions. We must thus gain a greater insight into the way in which a person interacts with a machine, particularly given that the interaction may be non-passive. We are primarily interested in the change in wrist and arm dynamics as the person co-contracts his/her muscles. It is observed that this leads to a change in stiffness that can push an actuated interface into a limit cycle. We use both experimental results gathered from a PHANToM haptic interface and a mathematical model to observe this effect. Results are relevant to the fields of rehabilitation and therapy robots, haptic interfaces, and telerobotics.

Background

There are several application areas where machines make an intimate contact with the user and in these situations it is important to gain a good understanding of human neuro-

musculo-skeletal dynamics. Several areas in the field of rehabilitation robotics require this type of close contact with a person and in these situations it is possible that some useful information can be gained from that contact. Close contact robots in rehabilitation include power-assisted orthotic mechanisms [1], robots in physical therapy[2,3], and EPP based telerobotics[4]. In non-rehabilitation applications, close contact robots are common in haptic interfaces and telerobotics.

To aid the design of close contact machines requires good knowledge of the human under conditions similar to those that will be experienced in practice. Although it is attractive to develop linear approximations of human dynamics as this allows for easier stability analysis, human arm dynamics are inherently non-linear and time dependent and include factors such as fatigue, posture, and movement history. In rehabilitation the clinical condition gives a further complication adding additional factors to the equation such as tremor, muscle atrophy, and limb flaccidity.

We use a two level approach to understanding human neuro-musculo-skeletal dynamics and investigate co-contraction in the process. An

experimental method allows in-situ data to be gathered at the first level. At a second level individual physiological elements in the joint of interest can be modelled and the composite dynamics then simulated.

Human System Identification

Several studies base a human system model on a second order mass, spring, damper approximation at the mechanical properties of various joints [5,6,7]. Standard techniques then allow the lumped characteristics of the human arm to be determined by applying a perturbing force, and then examining the positional response. A force feedback device such as the PHANToM (Sensable Technologies, Cambridge MA, USA) has the ability to both apply a force and measure the positional response of the user. The PHANToM was used in the following experiments and consists of a low impedance, 3 degrees-of-freedom, revolute manipulator where the traditional end effector is replaced by a thimble, through which the user interact with the device.

The workspace of the PHANToM is designed for movements of the finger and wrist, therefore it is these joints that will be the focus of the modelling. Previous studies of the impedance presented by the index finger [5] report several trends:

- There was little inter-subject variation in mass estimates.
- There was an approximately linear increase in stiffness with applied

force.

- There was a relatively large, near critically damped value of the damping ratio for fast transients.

In a study of the stiffness of the human wrist [7], the relationship between the angular position and the torque was modelled by an underdamped second order parametric model.

Experimental Method

Preliminary experiments were performed in order to assess the feasibility of developing mechanical impedance models for the human wrist and the metacarpal-phalangeal joint of the finger. The subject's elbow and other relevant joints were firmly secured via a splint so that the only movement was the joint being examined. The finger splints were rigidly attached to the tip of the PHANToM, via the thimble provided. Perturbations were applied via the base motor of the device of an amplitude determined by sampling from a normal distribution of zero mean, with a fixed period of 0.1s. The subject was either asked to relax, or to co-contract the appropriate muscles in order to oppose the motion. The subsequent displacement of the corresponding joint on the PHANToM was recorded.

Results and Analysis

The resultant positional output and estimated torque input data was used to construct a second order discrete time ARMA model relating the two variables. Such a model can then be converted to a second order mass-

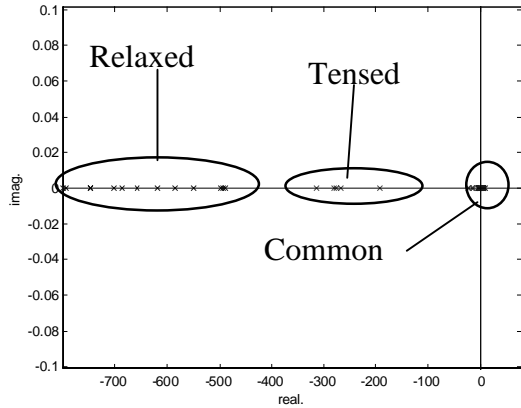


Figure 1: Continuous Time Model Poles for Human Wrist over 0.1s Time Window

spring-damper model of impedance in the continuous time domain, providing some estimate of the mechanical parameters of the impedance presented by the user. The plot in figure 1 illustrates the poles of the continuous time models for tensed and relaxed wrists. The data was analysed over a 1 second time window, taken from the beginning of the first step in torque.

A visual analysis of the data suggests three different regions for the location of poles, indicated on the diagram. The region near the origin includes poles for both contracted and relaxed conditions and is common throughout all the models developed. The poles are close to the origin, suggesting an unbounded position response to a step input. Several poles were unstable, which is an unrealistic suggestion, however, it is inferred that over a small displacement, away from the limits of movement of the joint, a suitable model for the impedance of the wrist is:

$$\frac{\theta}{T} = \frac{K}{s(s\tau(u) + 1)} \quad (1)$$

where K is the d.c. compliance, T and θ are the applied torque and resulting angular perturbation.

The unstable poles result from a lack of information present in the data, due to the long time constant of the wrist. Modelling over a longer time period may eliminate the instability. The model suggested in equation (1) is a gross oversimplification of the dynamic properties of the human wrist. However, it is reasonable to suggest that it does approximate the dominant mechanical properties of the joint over a limited displacement not approaching the limits of the joint's motion, prior to onset of sensory feedback or reflex actions. The time constant, τ , here depends on level of muscle co-contraction and many other factors, as indicated by the regions on figure 1. For low levels of muscle activation, the second pole of the system is in the 'Relaxed' region, further into the left hand plane, indicating a faster response time. With muscle co-contraction, the second pole of the system is shifted towards the origin in to the 'Tensed' region, indicating an increase in the stiffness. Results for the response of the finger to perturbations displayed similar behaviour.

As with varying levels of muscle contraction, three distinct regions are again evident in the pole placement. The model expressed in equation (1) is again applicable to the results, with τ being a function of input force. Region 1 represents the pole at the origin. Regions 2 and 3 display the variation in the mechanical parameters of the

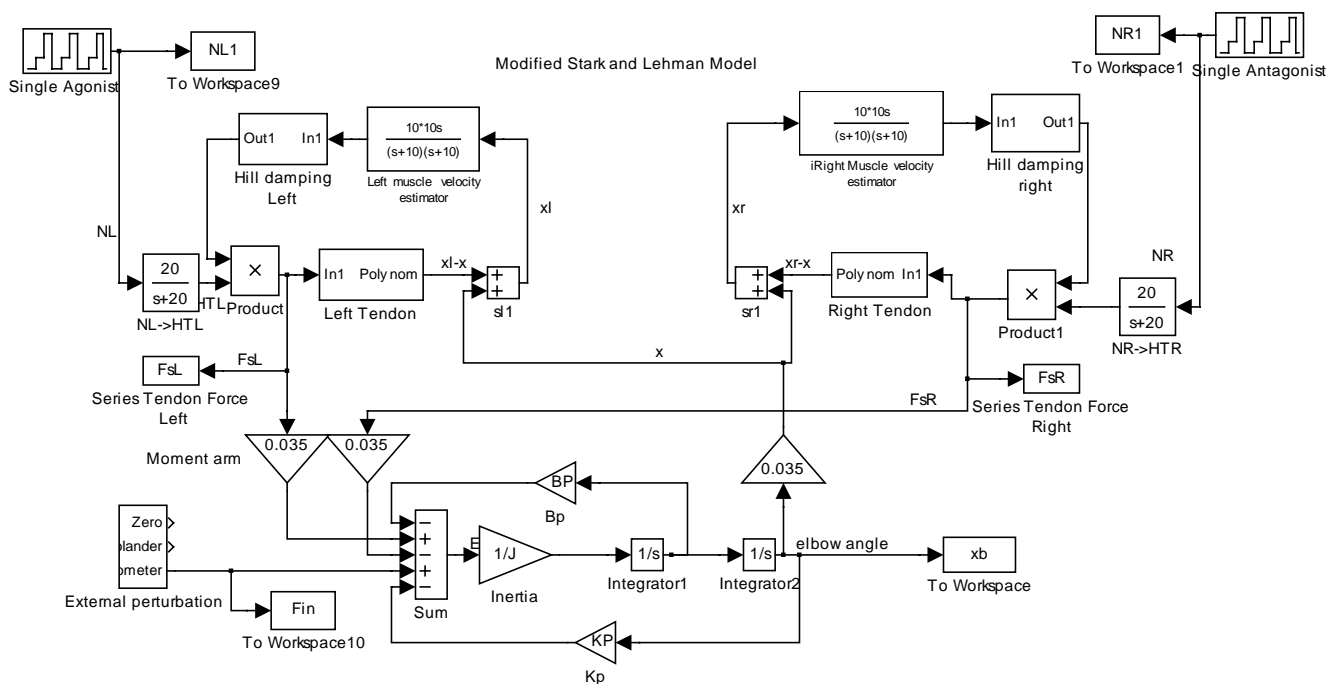
system with the magnitude of the perturbations. This indicates an increase in response time, and, hence, stiffness with increasing force, which agrees with the results presented by Haijan and Howe[5].

Simulation of co-contraction in the elbow

A non-linear elbow model has been developed, based principally on that of Stark and others [8] but adapting parameters from Prochazka[9] and Gossett[10]. This model is used to identify the elements that cause an increase in stiffness when agonist and antagonist muscles co-contrast. It is hypothesised that there are three mechanisms that contribute to the increase in stiffness when a person co-contraction their muscles

1. The Hill effect causes a drop in force in the shortening muscle, whereas the extending muscle exerts a larger force thus tending to restore the limb following a perturbation.
2. The non-linear length-tension relationship of the series tendon operates higher up the non-linearity when muscles are co-contracted thus causing a greater stiffness.
3. The reflex action of the golgi tendon organ.

The simulations done here illustrate the first of these and show that a non-linear series elasticity prevents high frequency vibration at high levels of muscle tension. This mechanism does not appear to contribute significantly to the increase of stiffness as muscles co-contrast. The third mechanism is currently unexplored.



Description of simulation

The simulation is shown in figure 2. A bimuscle model is used and the force of contraction is estimated by scaling the Hill damping hyperbola. The form of the Hill equation for contracting muscle is

$$\frac{F}{F_{act}} = 1 + \frac{(1 + a \text{fact})v}{Bh - v}$$

where $Bh = |V_{max}| a \text{fact}$. v is the muscle contraction velocity, F the force of contraction, F_{act} is a measure of muscle activation, and $a \text{fact}$ and Bh are the Hill constants. A cubic spline, with continuous first and second differentials at $v=0$, is used when the muscle is being extended. A shaping parameter $p=0.2$ is used to force an intercept on the positive x axis at $|V_{max}| p$. The velocity of the muscle with respect to the bone is estimated from position using a simple second order filter with a double pole giving a 3dB cut off at 5 rad/s.

The series tendon connecting the muscle to the bone, is modelled either as a linear element $F=Ke x$ or as a fourth power $F=Kx^4$. The spring constant in the latter case is adapted to fit data published by Evens and Barbernel [11] for the human palmaris tendon.

Table 1 shows values for other parameters along with comparison with other simulation studies. It should be noted that the tendons are assumed to translate force into torque via a constant moment arm, and gravitational effects are ignored.

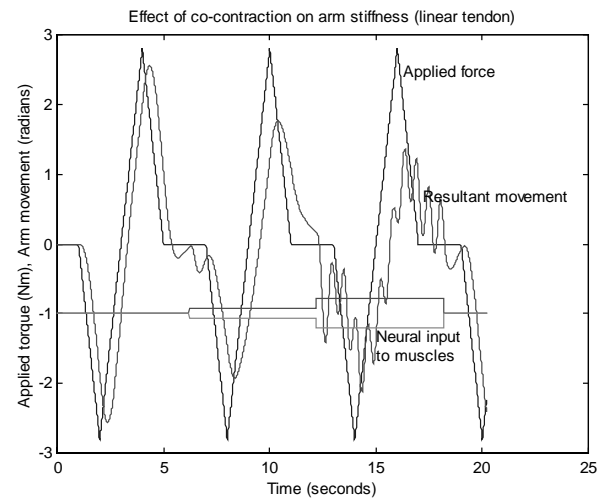


Figure 3: Stiffness of model with linear tendon.

Simulation results

Results of the simulation where the tendon is modelled as a linear spring are shown in figure 3. The applied torque is ramped down and then up to $+ 2.8$ Nm, and the resulting movement of the arm observed. When there is no co-contraction as indicated for the first 6 seconds, the elbow acts as a weak spring, with a small lag. Between 6 and 12 seconds the muscles are activated at about half their full strength. During

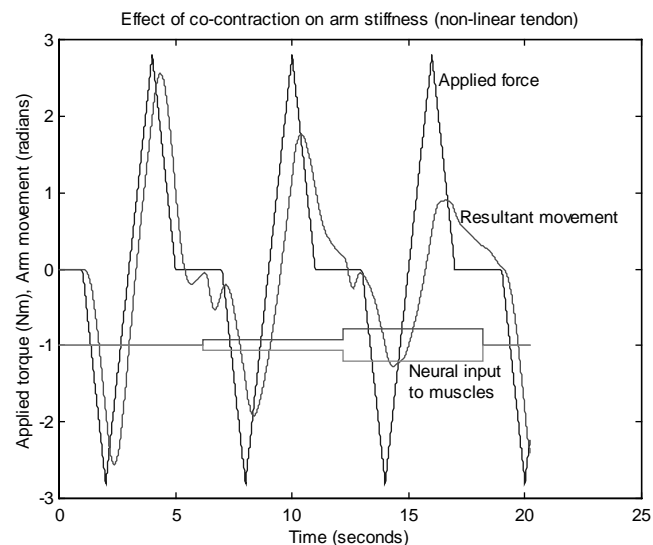


Figure 4: Stiffness of model with non-linear tendon

this period the stiffness does increase by a small amount, as can be observed by the change to the gradient of the position, and the lower movement peaks. At high levels of co-contraction a high frequency limit cycle is induced. Figure 4 shows the results when the tendon model is replaced by the non-linear equation $F=Kx^4$. Results are similar to those shown in figure 3 with possibly slightly more change in stiffness as muscles co-contrast. It is noted that the non-linear tendon suppresses the limit cycle observed in at high levels of co-contraction in the linear tendon model, this could be an artifact of the numerical integrator. It is somewhat surprising that the non-linear tendons do not contribute more to the change of joint stiffness observed in practice.

Discussion

System identification techniques are able to identify locally linear models for a person interacting with an

actuated interface as has been illustrated for the wrist data given. The model gives an adequate description but only for small movements away from the joint limits. The measurements of force and position were derived entirely from access to internal control parameters of the PHANToM and a model of its dynamics. Better measurements from the PHANToM would possibly improve the model estimates. However this demonstrates the potential of in-situ human model identification.

The danger of the more detailed non-linear physiological model is that it is sensitive to the choice of parameters for which there is little practical data. In addition the current model does not include a reflex neural circuit thus omitting a factor that undoubtedly has an influence on the change of stiffness as antagonist muscles co-contrast. However if a physiologically appropriate and accurate model can be developed from interaction data it can

	Gos94 Elbow/ forearm	Stark Neck/ head	Prochazk97 Cat solenus	This simulation Elbow/ forearm	
<i>J</i>	0.0772	0.0103		0.07	kgm ²
<i>m</i>	1.77		1		kg
<i>l</i>	0.177				m
<i>KP</i>	1	0.115	0	1	Nm/rad
<i>BP</i>	1	0.115	0	0.3	Nms/rad
<i>Ke</i> (tendon extensor)	0.1	2.29	20,000	7 (2000 N/m at 0.0035m)	Nm/rad Nm
<i>Kf</i> (tendon flexor)	0.1				Nm/rad
<i>Be / Bf</i>	1	Hill B	Hill B	Hill B	Nms/rad
<i>Hill af</i>		.25		.25	
<i>Hill Bh (=af Vmax)</i>		1.5	1	0.66	m/s

then be linearised for control system design or a simplified version can be used for model reference control techniques.

Conclusion.

Both experimental and simulation models of the human wrist and elbow have been discussed with advantages and disadvantages of each discussed. As in many areas it demonstrates the trade-off that must be made between simplicity and accuracy

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