Quantification of shoulder and elbow passive moments in the sagittal plane using a robot

Marko Munih*, Timotej Kodek

University of Ljubljana, Faculty of electrical engineering, Tržaška 25, 1000 Ljubljana, Slovenia

Introduction

The electrical stimulation of muscles is used often for exercise purposes. The user benefits this way with increased muscle strength and better overall movement control. There is also a positive effect on joint range of motion and, i.e. passive moments in joints that are normally smaller in-between and increase significantly toward both extreme joint rotations. Quantification of these (passive moments) is the topic of the presented study. The passive moments exerted in the human musculoskeletal system are an internal property of every joint in the upper and lower extremities. They arise mostly from the presence and deformations of structures such as tendons, ligaments, skin, joint capsules, inactive muscles and bones composing a particular joint. They could be expressed in terms of elastic and dissipative contributions.

There have been a large number of studies dealing with these properties, concentrated mainly on lower extremities. Some tried to capture properties for one joint, while others attempted to construct a model expressing the passive moments as a function of the two adjacent joint angles. Most have used a technique proposed by Audu and Davy [1], where this function was taken to be a double exponential curve, indicating a significant torque increase at extreme angles. Hatze [2] proposed a complex model that was further simplified into a hyperbolic curve, requiring an identification of a total of 53 elastic and viscous parameters for each degree of freedom in the human elbow.

A number of studies concentrated on the arm dynamics in the presence of a voluntary movement. Bennett and Hollerbach et al. [3] devised an ensemble parametric method for identifying the time-varying compliance of the human elbow joint, using an airjet actuator apparatus. Further studies by Xu and Hollerbach [4] on the elbow joint mechanical properties concentrated on estimating elasticity, viscosity and inertial contributions during a voluntary movement. In all these studies, the inertia contribution was shown to remain constant despite the varying voluntary muscle action, whereas elasticity and viscosity both increased and decreased proportionally with the applied muscle force. A number of other studies concentrated on the endpoint stiffness of the human arm mechanism. The studies of Engin et al., another study by Van der Helm, and Veeger et al. concentrated mostly on kinematics and also on passive resistive properties of the human shoulder complex. Unlike the work of Xu and Hollerbach, the study presented here is aimed at separating the effects of passive and active musculoskeletal contributions to the human arm dynamics.

This work concentrates on identifying the passive moments (i.e. elasticity, and dissipative effects) of the elbow and shoulder joints being moved one at a time through a large portion of their flexion-extension range in the sagittal plane. This was achieved by imposing slow i.e. static angular movements to a particular joint with a robot, while keeping the second joint at a fixed angle. The upper extremity was modelled in terms of an inverse dynamics equation for a three segment planar manipulator.

Methods

Mathematical modelling - In this experimental work the human arm was described as a three degree of freedom kinematic and dynamic structure. In Figure 1 the segment lengths, the masses and inertias, and the centers of gravity are depicted. These were obtained from the literature [5]. As in every other manipulator system, the
dynamic behaviour of the upper extremity, as a relationship between applied driving torques $\tau(u)$, environment forces $h$ and joint motion trajectories $q$ can be described as:

$$B(q)\ddot{q} + C(q, \dot{q})\dot{q} + G(q) + F_c,q + F_d \text{sgn} (\dot{q}) = \tau(u) - J^T(q)h.$$  

(1)

Here $q, \dot{q}, \text{ and } \ddot{q}$ represent the three component joint angle, angular velocity and angular acceleration vectors. The $B(q)$ represents the moments of inertia matrix, matrix $C(q, \dot{q})$ includes the centrifugal and Coreolis effects on the arm dynamics. The gravitational contribution is expressed with a three element column vector $G(q) = [g_1, g_2, g_3]^T$. The connection between the robot and the hand handle creates the closed kinematic linkage, with horizontal and vertical forces ($F_y,F_z$), and the moment around the axis perpendicular to the plane of motion ($M_x$). These forces are transformed to the joint space via a Jacobian transpose $J$. The joint muscle activity is expressed as $\tau(u)$, which is a function of muscle activity $u$. The viscous contribution in joints is expressed in terms of $F_c,\dot{q}$. $F_d \text{sgn}(\dot{q})$ indicates the dissipative (static friction) torques. The passive elastic torque contributions in a particular joint are expressed as $F_e,q$, with $F_e$ being a diagonal nonlinear matrix with elasticity coefficients for every single joint. Determining passive moments, as the sum of elastic and dissipative contributions, $F_e,q + F_d \text{sgn}(\dot{q})$, was the topic of this study.

![Figure 1 Model of the human arm, consisting of three segments](image)

The passive and static assumption - All measurement motions preprogrammed into the robot manipulator were slow, 0.3 rad/s for the elbow and 0.2 rad/s for the shoulder joint movement, and the angular accelerations values of up to 1.2 rad/s$^2$ at points of movement initiation. Therefore, assuming $\ddot{q} \approx 0$ and $\dot{q} \approx 0$ is well justified. From there, three terms $B(q)\ddot{q} \approx 0$, $C(q, \dot{q}) \approx 0$ and $F,\dot{q} \approx 0$ become negligible. The next assumption, $\tau(u) \approx 0$, is true if no voluntary muscle action whatsoever is induced by the subject. To verify and confirm this we used the EMG during a typical elbow flexion-extension trial. Used were surface electrodes placed on the four major flexion and extension muscles ie biceps long and short head, triceps and brachioradialis. All these assumptions shorten the original system (upper extremity) equation to:

$$F_c,q + F_d \text{sgn}(\dot{q}) = -G(q) - J^T(q)h.$$  

(2)
Starting from the right hand side: forces $h$ are measured from force cell that is mounted in-between the hand handle and the manipulator. $\mathcal{J}(q)$ is derived from kinematic data, similar is valid for the terms in $G(q)$, that are calculated through the mathematical model of the upper extremity. All these known terms result in a desired passive torque, represented with two terms on the left side of equation.

**Measurement** - During the measurement, the human arm held a bicycle-like circular rubber coated handle, mounted on the force cell and attached on the top of an industrial anthropomorphic 6-DOF robot, a Yaskawa MOTOMAN SK6. The force sensor was a 4 dimensional JR3. The maximum force for the specified output was $110 \, N$, with an acquisition resolution of 12 bits. The subject sat on a bus passenger seat, equipped with additional straps around the torso to prevent any body sway. The subject was asked to keep his muscles relaxed while holding the handle to fulfil the $\tau(u) \approx 0$ condition.

Two main sets of measurements were made:
- With the shoulder angle fixed at various angles, while the elbow angle was varied smoothly.
- With the elbow fixed, while the shoulder was moved through a range of angles.

In both cases the wrist was not fixed and was allowed to move freely, since the deviation from the neutral position was found to be only a few degrees. Before the particular measurements, ten different circular trajectories (not shown here) were programmed into the robot for each subject. The first five measurements concentrated on the elbow angle smooth variation from one boundary angle to the other and backwards, with the shoulder fixed at one of five angles $[-68^\circ, -40^\circ, 16^\circ, 10^\circ, 36^\circ]$. The shoulder angle was kept constant throughout the elbow movement only by programming an appropriate trajectory. The second set of trials focused on movements of the shoulder joint, with the elbow kept at constant angles $[20^\circ, 30^\circ, 40^\circ, 49^\circ, 59^\circ]$. In this case an orthosis was used for elbow angle fixation, which allowed angle adjustments from extension to a flexion angle of 85 degrees. The mass of the orthosis was considered in the model.

The 3D tracking system Optotrak was used to precisely record the movements during the experiment. The IR markers were attached to the skin above the rotation points of the three arm joints, to the handle and to robot manipulator joints to allow complete reconstruction of the measurement. All calculations mentioned, were performed off-line using Matlab, after the Optotrak and Force sensor data lowpass filtering at 5 Hz with a sixth order Butterworth filter.

Six healthy subjects were tested (body masses from 64 kg to 77 kg). They were all right-handed males, aged from 25 to 39 years. None had ever suffered from any kind of neuromuscular disease. Before the experiment, at least two preliminary movements were made to assure that the programmed trajectory was appropriate and that the subject was comfortable. After defining 10 different trajectories, a first measurement batch was made for the elbow, followed by the second ten trials for the shoulder joint. Beforehand, six twenty-trial sets were made on one particular subject (age 25, weigh 77 kg), with every set performed on a separate day. Every movement was repeated six times, for a total of sixty measurements. Hence, altogether six measurements were made for every movement i.e. extension to flexion and backwards. All other subjects were only measured twice for every movement.

**Results**

*Passive moment results for one subject* – Here, six repetitions for each curve in Figure 2 were made. It has to be noted that for these measurements the shoulder and elbow were not moved throughout their complete range of motion, because of a limitation imposed by the working space of the robot manipulator. Due to that, the
exponential nature of the passive moments for intact population, which is normally more expressed near the articular boundaries, is not very well evident. After the measurement itself, the static and kinematic data were used for calculation in equation (2), and resulted in passive moments given in Figure 2. Clearly, the passive moments of the shoulder are much less influenced by adjacent angle fixation than the ones of the elbow. The standard deviations (not shown here) for these traces are quite large, contributing to a large relative error at points where passive moments are close to zero. This error comes mostly from large force standard deviations. The curves hysteresis is arising from the muscle dissipative effects \( F_d \, \text{sgn}(\dot{q}) \), where the upper part of the curve always indicates movements from extension to flexion.

**Passive moment results for six subjects** - The same data analysis was used for the six other subjects in the study, with all the measurements made under the same conditions, each repeated twice. For practical reasons, the shoulder and elbow angles were not fixed completely equally for all subjects. This is mostly due to a fairly complex process of trajectory programming and different arm geometry among subjects and consequently in slightly different passive moments. Observed were similar standard deviations as earlier with one subject.

![Figure 2](image)

**Figure 2** Average elbow (a) and shoulder (b) passive moments as a function of both angles for the same person.

**Discussion and conclusion**

A new method for estimating arm passive moments is shown. While examining the data, repeatability is an important issue. The angle data is very repeatable, while the force sensor data on the other hand, shows more deviations. This is caused by a difficulty with which a subject is capable of maintaining the arm-robot connection fully equally in two successive trials. When comparing the outcome results between days it can be observed that every curve obtained shows a similar pattern. The force and kinematic data among various subjects show larger deviations than for one subject, originating basically from geometrical and dynamical differences (not shown here). This also explains why there is no straightforward correlation in the passive moments among all subjects. From data it can be recognized that the passive moments are strongly influenced by adjacent joint fixation, being much less evident for the shoulder joint as for the elbow. The reason lies in passive one and two-joint muscles, which span over both joints and are very likely the major contributor to the passive properties.

Another source of error is the term \( G(q) \). The values for segment masses and centers of gravity were at the time of the study estimated from the literature. It is expected that measured values of associated masses null the errors.
in next study iteration. The effect of length parameter errors in term $G(q)$ compensates, owing to Jacobian $J(q)$ with appropriate term in $J(q)h$. Hence, the error imposed by a marker misalignment is small and negligible.

In similar works the passive elastic torque was found to resemble a symmetrical double exponential curve with highly positive values at complete extension and negative ones at extreme joint flexion. It has to be underlined that the flexion-extension movement limits in this study never reached the articular boundaries of either the elbow or the shoulder joint due to a limited robot workspace and large working space of the arm. Therefore, the passive moments were here quantified only in the central region of the movement range.

In conclusion, the study presented here simultaneously determines all three passive moments in the human arm joints from the inverse dynamics model by using a robot manipulator. If compared to other studies on passive moments, the method followed here seems to be elegant from the subject point of view, with comparably less physical constraining of particular arm segments required. The experimental results shown here were obtained for healthy individuals, while we expect that impaired subjects in the study continuation will show values that will be clearly distinguishable. Such patients are considered to be good candidates for treatment with new rehabilitation treatment devices such as haptic robots only, or in combination with electrical stimulation.

References


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