

Identifying shoulder and elbow passive moments and muscle contributions during static flexion-extension movements in the sagittal plane

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Abstract

The main goal of this study was an assessment of the shoulder and elbow joint passive moments in the sagittal plane. An industrial robot was used for moving a passive arm very slowly from extension to flexion and backwards. The motion was constrained to the sagittal plane, with a three degree of freedom planar structure assumed for the human arm. Comparing the obtained passive moments of six young male subjects showed a large adjacent angle dependency. Later on, voluntary muscle joint torques for one particular subject and trial were calculated, based on the acquired passive moment data. The presented methodology is aimed at an application on a rehabilitation robot.

1 Introduction

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 [1, 2] composing a particular joint. They could be expressed in terms of elastic and dissipative contributions [2]. Several authors in the past concentrated only on the elastic effects [1, 3]. There have been a large number of studies dealing with these properties, out of which the majority were concentrated on lower extremities [3–5]. In addition to examining torque-angle properties for one joint, many authors have attempted to construct a model expressing the passive moments as a function of the two adjacent joint angles. Most [2, 3] have used a technique proposed by Audu and Davy [6] where this function was taken to be a double exponential curve, indicating a significant torque increase at extreme angles. On the other hand, Hatze [1] proposed a model, consisting of a sum of several individual tissue exponential contributions relating to an observed joint. This relation was further simplified into a hyperbolic one, requiring an identification of a total of 53 elastic and viscous parameters for each

degree of freedom in the human elbow joint (i.e. flexion-extension and pronation-supination). It has to be pointed out that all these studies were made without any voluntary muscle action.

There has also been a number of studies concentrating on the arm dynamics in the presence of a voluntary movement, particularly in the elbow joint. Following a study on torques produced in the elbow joint with voluntary movements [7], Bennett and Hollerbach *et al.* [8] 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 [9, 10] on the elbow joint mechanical properties concentrated on estimating elasticity, viscosity and inertial contributions during a voluntary movement, using a similar technique and a two-dimensional device capable of imposing random torque perturbations. 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 comprehensive analysis of the kinematic and dynamic behaviour of the shoulder mechanism providing a good insight into mechanics of the shoulder mechanism, was presented by Van der Helm [11]. Some parameters acquired in the study of Veeger *et al.* [12] were also a good lead to our study.

Unlike the work of Xu and Hollerbach [9, 10], the study presented here is aimed at separating the effects of passive and active musculoskeletal contributions to the human arm dynamics. This work firstly concentrates on identifying the passive moments (i.e. elasticity, and dissipative effects) of the elbow and shoulder joints being robotically 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, while keeping the second joint at a fixed angle. The upper extremity was modeled in terms of an inverse dynamics equation for a three seg-

ment planar manipulator [13, 14]. After determining the passive moments for a particular subject, an active torque contribution produced by muscles was measured for comparison reasons. The subject produced desired voluntary muscle action, to verify the torque amplitude in this particular setup.

2 Methods

Mathematical modeling

In this experimental work the human arm was described as a three degree of freedom kinematic and dynamic structure (Fig. 1). The segment lengths are denoted with a_i , their centers of mass with l_i while q_i indicate the positive angle directions with respect to the zero position (*dashed line*). The masses and inertias are presented with the m_i and I_i variables. The centers of gravity were expressed as a distal length from the joint marked with the same index.

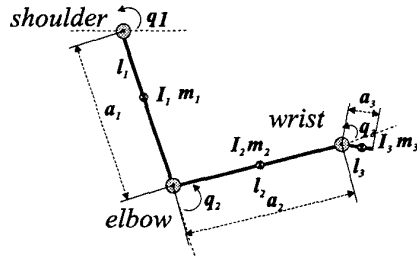


Figure 1: Geometric definitions for the assumed human arm structure, consisting of three segments.

As in every other manipulator system, the dynamic behaviour, as a relationship between applied driving torques $\tau(u)$, environment forces h and joint motion trajectories \ddot{q}, \dot{q}, q of mechanical joints can be described as [14]:

$$\begin{aligned} B(q)\ddot{q} + C(q, \dot{q})\dot{q} + G(q) + \\ + F_v\dot{q} + F_e q + F_d \text{sgn}(\dot{q}) = \\ = \tau(u) - J^T(q)h \end{aligned} \quad (1)$$

Here q, \dot{q} and \ddot{q} represent the joint angle, angular velocity and angular acceleration vectors, which are functions of time, but were for simplicity reasons denoted as q instead of $q(t)$. They can be expressed as column vectors with indices 1, 2 and 3 referring to the shoulder, elbow and wrist respectively.

The *gravitational* contribution is expressed with a three element column vector, where every element g_i represents the moment generated at the joint i axis due to the presence of gravity:

$$G(q) = [g_1 \quad g_2 \quad g_3]^T, \quad (2)$$

where

$$\begin{aligned} g_1 &= g_0 \{ [l_1 m_1 + a_1(m_2 + m_3)]c_1 + (l_2 m_2 + a_2 m_3)c_{12} + l_3 m_3 c_{123} \}, \\ g_2 &= g_0 [(l_2 m_2 + a_2 m_3)c_{12} + l_3 m_3 c_{123}], \\ g_3 &= g_0 l_3 m_3 c_{123}. \end{aligned}$$

In this equation the cosines were simply denoted as: $c_1 = \cos(q_1)$, $c_{12} = \cos(q_1 + q_2)$ and $c_{123} = \cos(q_1 + q_2 + q_3)$. While the individual segment lengths a_i for a particular person were determined from IR markers used by a 3D positioning system, the masses m_i and gravity centers l_i , were obtained from the literature [15]. The gravitational acceleration g_0 was taken to be 9.81 m/s^2 .

The connection between the hand and the robot handle (see section 3) creates a closed chain kinematic linkage. Thus, the end effector connection is described as a three dimensional vector with its vertical and horizontal forces (F_y, F_x) and the moment around the axis perpendicular to the plane of motion (M_x).

These forces have to be transformed to the joint level with the Jacobian matrix $J^T(q)$ as seen in Eq. 1. The joint muscle activity is expressed in terms of the active contribution $\tau(u)$, which is a function of muscle activation u .

The viscous contribution of the system is expressed in terms of $F_v \dot{q}$. F_v is a 3×3 diagonal matrix of viscosity coefficients. $F_d \text{sgn}(\dot{q})$ indicates the *dissipative torques* with F_d being a 3×3 diagonal matrix. In the literature this product is usually denoted as the static friction torque [14]. Finally, the *passive elastic torque* contributions in a particular joint are expressed with the product $F_e q$, where F_e is a 3×3 diagonal matrix with the elements expressing the elasticity coefficients of 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.

The passive and static assumption

All measurement motions preprogrammed into the robot manipulator were slow, with an end effector constant speed of 0.1 m/s . The arm joint angular speeds did not exceed 0.3 rad/s for the elbow and 0.2 rad/s for the shoulder joint movement while the accelerations and decelerations were low. Due to that, the contributions of all terms in Eq. (1), relating to either joint velocities or accelerations were negligible:

$$B(q)\ddot{q} \approx 0, \quad C(q, \dot{q})\dot{q} \approx 0, \quad F_v\dot{q} \approx 0 \quad (3)$$

The next observation concerns the term $\tau(u)$ in Eq. (1). The subject was instructed before the experiment, to induce no voluntary muscle action, leading to a further assumption:

$$\tau(u) \approx 0 \quad (4)$$

To verify if this was justified, the EMG of a typical elbow flexion-extension was recorded prior to the large batch of experiments, to assess the difference between the EMG of the active muscle, compared to the one of the passive muscle. The surface electrodes were placed on the four major flexion and extension muscles by a skilled professional (*i.e.* *biceps* long and short head, *triceps* and *brachioradialis*). The measurement demonstrated that no EMG activity in those muscles contributing to the movement was present. To limit the space of presentation the EMG curves are not shown here.

All these assumptions were accounted for and considered in calculation of the passive elastic torques, in Eq. (1), modifying now to:

$$F_e q + F_d \text{sgn}(\dot{q}) = -G(q) - J^T(q)h \quad (5)$$

The passive torque represented with the left side of equation (5) is represented as a time and angle dependent column vector $\tau_p(q, t)$, or simply τ_p :

$$\tau_p = [\tau_{p1} \quad \tau_{p2} \quad \tau_{p3}]^T \quad (6)$$

Calculating the voluntary muscle contribution

After the series of passive moment experiments, the subject was asked to apply a voluntary force in the upward-downward direction. The manipulator motion trajectory was the same as with the passive movement experiment, meaning that the static assumption in Eq. (3) was still satisfied. After having the results from Eq. (5) and considering the passive and static assumptions (Eq. 4, Eq. 3), the active torques arising due to the voluntary muscle action can be expressed from Eq. (1) as:

$$\begin{aligned} \tau(u) &= F_e q + F_d \text{sgn}(\dot{q}) + G(q) + J^T(q)h = \\ &= [\tau_1 \quad \tau_2 \quad \tau_3]^T \end{aligned} \quad (7)$$

This three-dimensional column vector represents voluntarily produced muscle torques in observed joints.

3 Measurement

In the performed experiment a positionally controlled anthropomorphic 6-DOF industrial robot (*Yaskawa*® *MOTOMAN sk6*) was used for imposing a slow linear circular movement on the human arm in the sagittal plane (Fig. 2). A *JR3*® 4 dimensional strain gauge force sensor was mounted on the manipulator end effector and used for force data collection. The maximum force for the specified output was ± 110 N, with an acquisition resolution of 12 bits. A bicycle-like circular rubber coated handle was mounted on top of the sensor in such a way, that rotation around the *x* axis was freely allowed. The next element in the system was a bus passenger seat, equipped with additional straps as evident from Fig. 2. The plane of motion was perpendicular to the ground and

fully aligned with the sagittal plane of the subject. In the first part of the experiment, the subject was asked to keep his muscles relaxed while holding the handle.

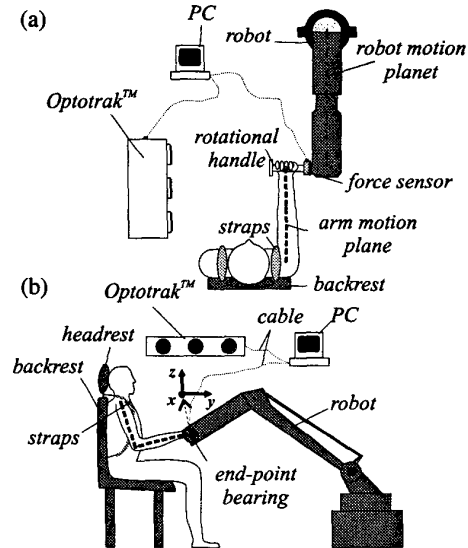


Figure 2: Experimental setup from above (a) and a side view (b).

The handle was held gently, while still allowing the arm to stay in good contact during the movement. Before starting the real measurements, it was also inspected whether the slight muscle activation due to gripping had any significant effect on the passive torque identification process. No significant difference was found when comparing this data to the one when the hand was tightly strapped to the handle.

Due to the free handle rotation the hand dynamic parameters of the hand were properly adjusted. By using the mass and all geometric dimensions of the handle which were accurately measured before the experiment. Two main sets of measurements were made:

1. With the shoulder angle fixed at various angles, while the elbow angle was varied smoothly.
2. 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 controller 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 different angles ($-68^\circ, -40^\circ, -16^\circ, -10^\circ, -36^\circ$). The shoulder angle was kept constant by programming an appropriate trajectory, using no additional fixation mechanisms.

The second set of trials focused on movements of the shoulder joint, with the elbow kept at constant angles ($20^\circ, 30^\circ, 41^\circ, 49^\circ, 59^\circ$). For fixating the elbow angle, an orthosis was used, which allowed angle adjustments from complete extension to a flexion angle of 85 degrees.

The mass of the orthosis was included into the calculation of the $G(q)$ matrix, utilizing the adjusted upper and forearm masses and center of gravity locations.

The orthosis masses and centers of gravity were accurately determined before the experiment.

The 3D tracking system *Optotrak*® was used to record precisely the movement coordinates during the experiment. The IR markers were attached to the skin above the rotation points of the three arm joints in consideration, to the handle and to robot manipulator joints to allow for later verification and complete reconstruction of the measurement. All calculations mentioned here were performed off-line using Matlab®. The *Optotrak*® and Force sensor data were both lowpass filtered at 5 Hz using a sixth order Butterworth filter provided by the Matlab® Signal Processing toolbox.

Six healthy subjects were tested with body masses ranging 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. All were asked to sit in a chair, lightly grip the robot attached handle and not exert any voluntary muscle action. 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 set of the first ten measurements was made for the elbow and the second ten, for the shoulder joint, where every trajectory was recorded twice.

Initially, three twenty-trial sets were made on one particular subject (age 25, weigh 77 kg), with every set performed on a separate day. On every day each of the ten different movements was measured twice in a row. Every movement was repeated four more times on two different days, for a total of thirty measurement pairs. Hence, all together six measurements were made for the same movement (*i.e.* extension to flexion and backwards).

In the last part of the experiment, only one subject was instructed to exert voluntary muscle action into an upward and downward direction in order to calculate the active muscle torque contributions according to Eq. (7).

4 Results

The results section is composed of three parts. First, a detailed overview of data acquired for one person is given.

The second part includes measurements on six persons to gain insight into data variability among several persons. Thirdly, an experiment with active muscle torque for one subject is included for comparison reasons and to verify the torque amplitude measured in this setup.

Passive moment results for one subject

In total six measurements were made for the same movement (*i.e.* extension to flexion and backwards). In Fig. 3 average time courses for force and kinematic data of the six same movement trials for one configuration, where the elbow was moved through a large portion of the motion range are shown.

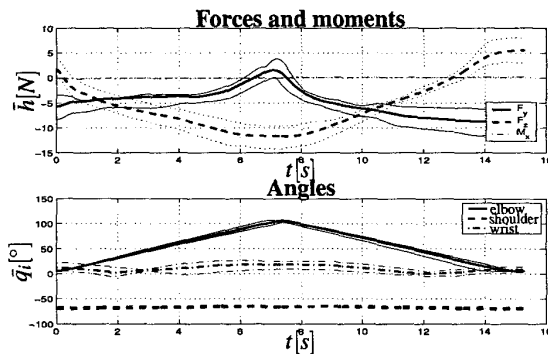


Figure 3: The average handle force \bar{h} and joint angle \bar{q}_i trajectories with their maximum and minimum six trial deviations for one subject, while the shoulder was fixed at $q_1 \approx -68^\circ$ and while the wrist angle variation was small.

The deviations of the \bar{h} and \bar{q}_i vectors were found to be within a range of ± 2.9 N and $\pm 14.5^\circ$ respectively. The x axis torque was negligible due to a bearing in the mechanism of the handle.

In Fig. 4 all ten computed average passive moments are presented as a function of the elbow and shoulder angles. The fixation angles of the elbow (q_2) and shoulder (q_1) as measured by the *Optotrak* system, are also denoted.

In all curves a hysteresis arising due to muscle dissipative effects can clearly be observed [2], where the upper part of the curve always indicates movement from extension to flexion. The hysteresis average is known to be the passive elastic moment, which was the interest of some particular studies [1, 3]. Curves are not of the same length, because the angle range was different for every particular movement.

Passive moment results for six subjects

The same data analysis was used for all six subjects in the study and all measurements were made under the same conditions. Again, every movement was measured twice for every subject.

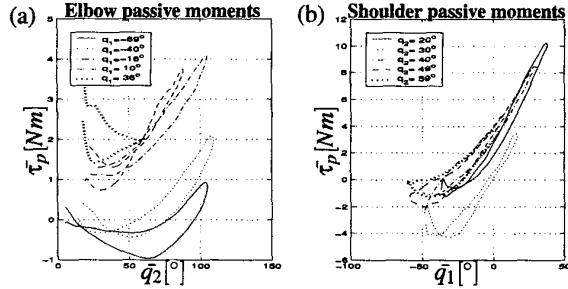


Figure 4: All five average elbow (a) and shoulder (b) passive moments as a function of both angles for the same person. Every curve represents an average of six measurements.

It needs to be emphasized that the shoulder and elbow angles were not fixed completely equally for all subjects, due to a fairly complex process of trajectory programming and different arm geometry among subjects. This fact inseparably results also in slightly different passive moments.

to limit the space of this presentation, the ten calculated passive moments for all subjects are not shown. To see the variation of results among all six subjects, only traces for one elbow and one shoulder trajectory configurations are shown (Fig. 5).

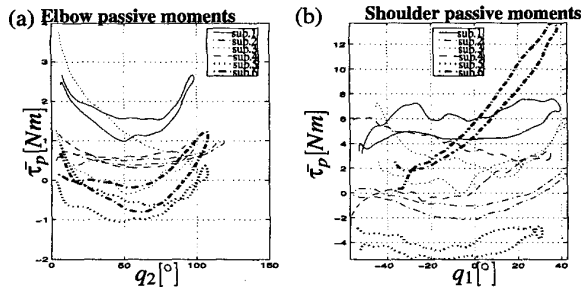


Figure 5: Obtained passive moments for all six subjects performing two particular movements when the shoulder joint was fixed at $\bar{q}_1 \approx -63^\circ$ (a) and the elbow joint at $\bar{q}_2 \approx 17^\circ$ (b).

Active muscle torque experiment

One subject (age 25, weigh 77 kg) was asked to exert a vertical upward force in the first trial and a downward direction force in the second trial. During this, his arm was moved through one of the trajectories for the elbow from extension to flexion and back to extension (Fig. 6). Based on the identified passive moments in these trials, the active moments $\tau(u)$ were computed according to Eq. (7).

The left side of Fig. 6 stands for the force exerted upward, which means that all the three joints had to act in

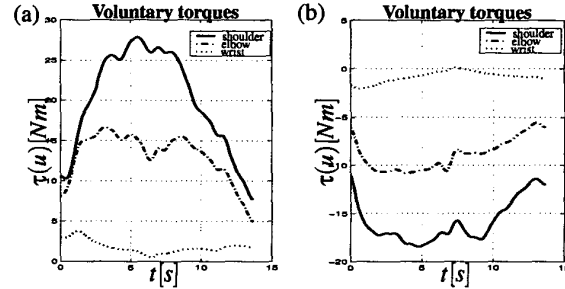


Figure 6: Computed voluntary torques during an upward (a) and downward (b) applied force.

extension. This explains the positive torque values acting in the counter-clockwise direction in accordance with the notation explained in Fig. 1. Because the downward pull action on the handle results in a flexion movement, the right part of Fig. 6 shows negative torque values. Both indicate weaker torques at flexion-extension edges, coinciding with the fact that the joints are unable of large torque generation in these angles.

This method enables the determination of voluntary torque contributions in all three joints on a certain predefined path. Before this can be made passive moments need to be determined from a passive movement experiment as explained before. With subjects suffering from neurological disorders, however, keeping the arm passive might not be feasible.

5 Discussion and conclusion

In this paper a method for estimating arm passive moments is proposed, which according to our knowledge, has not been used before. In the measurement process, one healthy individual was studied more in detail as described in section 3. The repeatability of data obtained from six measurements can be observed in Fig. 3. While the angle data is very repeatable, the force sensor data on the other hand, shows larger 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. Five elbow and five shoulder passive moments were inspected, with adjacent joints being fixed at various angles (Fig. 4).

Secondly, five more healthy subjects were measured in the same way. The force and kinematic data among subjects show larger deviations than for one subject, arising from geometrical and dynamical differences. This also explains why there is no straightforward correlation in the passive moments among all subjects (Fig. 5). A fairly large amplitude variation among different subjects was observed, especially in the shoulder joint.

Finally, the identified passive moments were used further to determine muscle produced joint torques for an elbow

flexion and extension trial (Fig. 6). As expected, the values are most prominent for the shoulder joint, which has the largest effect on the upward and downward movement of the arm.

It can be seen that the passive moments are strongly influenced by adjacent joint fixation. However, this is much less evident for the shoulder joint, as it is for the elbow (Fig. 4). It is also obvious that the shoulder passive moments are far larger than the ones obtained 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. While there are only seven muscles producing elbow joint movements, there are fifteen which are involved in the shoulder, with a total cross section area far greater than the one of the elbow muscles. Apart from this, the biceps and triceps muscles which contribute to elbow joint motions are two-joint muscles spanning the whole upper arm and hence influence the passive properties of both the shoulder and the elbow joint.

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. This occurs due to a rather limited robot workspace and almost no physical constraining of the arm. Therefore the passive moments were identified only in the central region of the movement range.

Apart from the relatively large force sensor data deviations (Fig. 3), another source of error is also the term $G(q)$ in Eq. (5) which was calculated by using the segment masses m_i , lengths a_i and centers of gravity l_i , from the literature [15]. Because the segment mass estimation m_i affects only the term $G(q)$ in this equation, the inexact value causes significant errors to the passive moment calculation. On the other hand, segment length and center of gravity location errors do not affect the result as much. The effect they have on the term $G(q)$ cancels itself with that from the environment contribution term $J^T(q)h$ (Eq. 5). The reason lies in the Jacobian matrix $J^T(q)$ which also depends on l_i and a_i . Hence, the error imposed by a marker misalignment, is not very prominent, resulting in low percentage changes in segment lengths a_i and subsequently centers of gravity l_i .

Because all experimental results here were obtained for healthy individuals, a further study of impaired subjects, is expected to be very beneficial. Passive moment patterns may prove to be much different than the ones found in healthy subjects. If the subject is not able to passively hold the arm this could be seen from the obtained results. The approach taken here may be useful in estimating what the muscle produced joint torques are. A future study would inspect the ability of the subject to exert joint torques in various directions and eventually even joint torques during some arm everyday movements without using the robot.

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