Identification of the Mechanical Impedance at the Human Finger Tip

Rapid transients were applied to the outstretched human index finger tip, which resulted in motion primarily as the metacarpophalangeal (MCP) joint in extension and in abduction. A second-order linear model was fit to apply angamply 20 milliseconds of the force and displacement data to determine the effective mechanical imped- ance at the finger tip in terms of mass, damping, and stiffness parameters. These parameters were estimated over a range of mean finger tip force (3–20 N for extension, 2–8 N for abduction). Effective translational finger tip mass for each subject was relatively constant for forces in excess of 3 N and for frequencies throughout the abduction trials. Stiffness increased linearly with muscle activation. The estimated damping ratio for extension trials was about 0.7, twice the ratio for abduction.

I Introduction

Mechanical impedance characterizes the relationship between input motion and externally applied input grip, and is comprised of both a static component relating forces and displacements, and a dynamic component relating forces to velocities and accelerations. Active modulation of limb im- pedance by the central nervous system is an essential part of effective motor control (Missiaen-Valdi et al., 1985; Horgan, 1990). Similarly, mechanical analyses and robotic experiments have demonstrated that appropriate selection of mechanical imped- ance facilitates the execution of control tasks (Winston, 1982; Asada and Arai, 1988).

Mechanical impedance is also important in human-machine interaction, and the design of effective haptic interfaces for teleoperated systems must consider the variability of human impedance. For example, both theoretical and experimental re- sults suggest that human operator impedance is a key factor in determining the stability and performance of haptic interfaces for teleoperated manipulation (Colgate and Brown, 1994). In the future, an understanding of human impedances and imped- ance-based control strategies could provide practical insights to solve problems in prosthetic limb design or in understanding neurovascular dysfunctions (Kramen and Hunter, 1982).

The primary goal of this study is to provide quantitative impedance measurements for haptic interface applications such as determining dynamic loads for mechanical design, and setting feedback gains for control system analysis. For these applica- tions the primary interest is in the effective impedance presented by the human hand to the mechanical interface, often in a relat- edly un-constrained posture. This is a different emphasis from many previous studies of human limb impedance (e.g., Agarwal and Gottlieb, 1977; Cowlishew et al., 1976; Joyce et al., 1974), where the goal was to identify the impedance of a partic- ular joint for the purpose of characterizing motor behavior. Previous human studies of mechanical impedance include single-joint measurements of the ankle (Hunter and Kramen, 1982), elbow (Jones and Hunter, 1990a; Bennett et al., 1992), and multijoint measurements of the arm (Hogan, 1990; Do- lan et al., 1993) and leg (McMahan, 1984). Human fingers may be expected to evidence a number of important differences with these systems. First, the skeletal kinematics and motola-

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II Methods

A Apparatus and Procedure. Five healthy subjects (one female; four male; ages 23–39) voluntarily participated in this study. Subjects grasped a rigid handle and rested their hand, wrist, and forearm on a rigid horizontal surface. The index finger was extended to press upon a force sensor attached to the piston of a pneumatic cylinder, which was rigidly clamped to a table (see Fig. 1). The experiment was conducted in two modes: One measured extension, and the other abduction of the index finger. The distance from the point of contact to the handle was individually set to account for finger length variations among subjects. The point of contact was the center of the finger pad for extension and the medial edge of the finger tip at the center of the distal phalanx for abduction. Subjects were instructed to increase finger force against the apparatus gradually (mean rate: N/s). As the subject exceeded a variable force threshold, a solenoid valve opened, causing air to rush into the cylinder. This caused the piston to rapidly displace the tip of the fully extended index finger approximately 5 mm. Because the resulting rotational motion occurred primarily at the MCP joint, we assume that the limb mass proximal to the joint (the palm, wrist, and arm) can be considered a mechanical ground.

For extension testing, eight trials were conducted at each of six threshold levels (2–4–6–8–12–20 N) for each of the five subjects. No effects of order were observed on the identified parameters (see Results, below). For some of the subjects, 20 N approached the maximum voluntary force level in the prescribed configuration. After four of the eight trials at a given force threshold level, the threshold level was changed with the intent of minimizing anticipation as well as limiting fatigue; the order was four trials each at 4–2–12–6–8–20–8–20–6–12–2–4 N. Subjects were also permitted to pause between sets of four trials, as long as hand position was not altered until all trials were completed.

For abduction testing, eight trials were conducted at each of four threshold levels (2–4–6–8 N) for each subject. For most subjects, 8 N approached the maximum voluntary force level attainable in abduction-adduction. Tests were conducted in a similar fashion to the extension trials, with the order of force threshold levels 4–6–2–8–6–2–4–8 N.

Several additional experiments were conducted for comparative purposes, as described below. To characterize the role of force feedback compliance at low forces, a force threshold of 1 N was noted for each experimental mode. Similarly, to identify the effects of interphalangeal (IP) joint motion, a rigid splint was also affixed to the finger of two subjects for additional extension testing at five of the six force levels (1–2–4–6–12 N). The wooden splint was tightly bound to the index finger to compress the finger pad and prevent motion at either IP joint.

A piezoelectric accelerometer and a force sensor (with sensitivities of 20 kHz and 70 kHz, respectively) sensed the finger tip force and acceleration. These signals were sampled at 20 kHz with a 12-bit analog-to-digital converter. The measured acceleration was numerically integrated to generate the velocity and position data for the finger tip.

To accommodate variation in finger force threshold levels (3 N–20 N), the air pressure at the valve was varied from 240 kPa at low finger tip force levels to 340 kPa for the trials at the highest level. The set time for the end plate of the piston to travel through its fixed 5 mm displacement varied between 14 and 20 milliseconds. The slight voluntary ranging of the baseline force level during the course of data acquisition was subtracted from the force record by measuring the slope of the force ramp (average rate was approximately 5 N/s) prior to the onset of subject motion.

B Model and Fitting Technique. A linear second-order model

\[
mx(t) + bx(t) + kx(t) = f(t)
\]

is assumed to represent the translational relationship between applied force \( f(t) \) and resulting displacement \( x(t) \), velocity \( kx(t) \), and acceleration \( m \dot{k}x(t) \) of the index finger tip. The parameters \( m \) and \( b \) represent the effective mass (kg), \( b \) the viscous damping (N/m/s), and \( k \) the stiffness (N/m) at the tip. Note that this lumped element translational model referred to the finger tip can be easily converted to a lumped element rotational analog about the MCP joint (Coehn and Plass, 1990) using the subjects’ finger lengths (Subjects 1–5, in meters: 0.102, 0.096, 0.101, 0.090, 0.097). For purposes of applying our findings to the design of haptic interfaces for teleoperation (e.g., Howes, 1992), we are specifically interested in the effective finger tip translational parameters.

The applied force \( f(t) \) and finger tip acceleration \( kx(t) \) are measured with respect to an assumed zero baseline immediately prior to cylinder expansion. Only changes from this baseline affect the estimated parameters. Velocity \( kx(t) \) and displacement \( x(t) \) are similarly defined from a zero baseline. Equation (1) can be written in matrix notation as a discrete system:

\[
\mathbf{M} \ddot{\mathbf{x}} + \mathbf{B} \dot{\mathbf{x}} + \mathbf{K} \mathbf{x} = \mathbf{F}(t)
\]

where \( \mathbf{F}(t) \) denotes a vector of discrete sampled values corresponding to the force transient, and \( \mathbf{K}, \mathbf{M}, \mathbf{B} \) an \( n \times 3 \) matrix of the resulting motion variables. Determination of the parameter values \( m, b, \) and \( k \) is accomplished by the division of the matrix \( \mathbf{K}, \mathbf{M}, \mathbf{B} \) by the force vector \( \mathbf{F}(t) \) to give a single least-squared error fit using the MATLAB

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software package. In addition, the damping ratio, defined for a second-order system as

$$\zeta = \frac{b}{2\sqrt{mk}} \quad (3)$$

is computed from the m, k, and \( \xi \) estimates.

To confirm the operation of the apparatus and fitting technique, the system was initially tested by expanding the cylinder with no contact while varying the height of the endplate (5 g to 20 g), and against three different springs of known stiffness (200 N/m to 700 N/m). The measured inertial mass was always estimated to be within 2.5 percent. This procedure was also used to identify both the effective driving mass of the apparatus and of the finger splint described above, each of which was subtracted from the total exciting mass estimated from subjects' trials.

In preliminary experiments to confirm the invariance of the identification technique with changes in input, trials were conducted with the pneumatic pressure halved and total cylinder displacement diminished first by 1/2 and then by 1/3. This resulted in corresponding changes in input force levels and waveform shapes. The consequent mean identified impedance parameters at each force level were within one standard deviation of those found with the input used throughout the trials reported below, confirming the insensitivity of impedance estimates to variation in input waveform shape.

III Results

Data from a typical trial are shown in Fig. 2. Each trial begins with large peaks in both force and acceleration, indicating high inertial forces at the initial motion of the piston. The acceleration becomes negative and the velocity diminishes after approximately 10 milliseconds, due to added pneumatic resistance at the exhaust port of the cylinder because of compression of the air in the cylinder above the piston. The displacement rises steadily after the initial acceleration. The detailed shape of individual acceleration and force waveforms varied from trial to trial because of the dynamics of the pneumatic actuator. However, the preliminary experiments described above showed that even large variations in the shapes of the waveforms do not produce significant variation in impedance parameter estimates.

To illustrate the ability of the model and the fitting procedure to account for the observed behavior, a comparison of the measured and calculated forces for a typical trial is given in Fig. 3. The calculated force is computed by multiplying the kinematic matrix \([N, k, x]\) by the estimated parameter value vector \([m, b, k]\). Also shown in Fig. 3 are the inertial (red), damping (blue), and stiffness (green) force components that comprise the calculated force. Note that during the first few milliseconds of expansion the inertial force dominates, while in the latter part, the damping and stiffness forces become significant. The variance accounted for (VAF) by the model quantifies the quality of fit of the model (Jones and Hunter, 1990b); in this trial, the VAF is 94 percent, and the average VAF throughout the experiment is 97 percent.

The variation with force threshold of mass, damping, stiffness, and damping ratio for one subject's extension trials is shown in Fig. 4. Estimated parameter mean values and standard deviations are depicted at each force threshold level. All five subjects show similar steadily increasing damping and stiffness parameters with increasing force threshold, and a mass estimate that is relatively constant or rises so a constant plateau at force levels greater than 6 N.

Figure 5 and Table 1 show the estimated parameters for extension for all five subjects. Effective mass estimates ranged from 2.7 g to 6.7 g at 2 N, and 5.1 g to 6.7 g at 20 N. Both damping and stiffness increase nearly linearly with force threshold, although damping has a large extrapolated zero-force value, while stiffness has a near-zero value. Damping nearly

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doubled across the force range, rising from a mean of 2.2 N/m to a mean of 4.0 N/m, while stiffness rose from a mean of 206 N/m to 800 N/m across the 2 N to 20 N range. Subjects were nearly critically damped (ζ ≈ 1) at force levels greater than 4 N in the mean, s, at each force level ranged between 0.92 and 1.15.

Auscultation of the transient by the subjects was not a significant factor in the results. The difference in parameters estimated in the first set of four trials compared to those estimated in the second set across all force levels was not significant (p > 0.25).

Similarly, sequences of trials for all subjects showed no training effect. There was no significant difference in parameters estimated in the first of a set of four trials as compared with parameters estimated in the other three trials (p > 0.25).

Figure 6 and Table 2 show parameter estimates for index finger abduction across the range of finger tip force thresholds tested (2–8 N). Mass estimates were relatively constant, ranging from 4.8 g to 7.0 g at 2 N and 8 N. Damping and stiffness estimates showed similar qualitative trends as in extension and rose with finger tip force. Damping increased from a mean of 1.84 N/m at 2 N to a mean of 2.72 N/m at 8 N, while stiffness increased from 230 N/m at 2 N to 520 N/m at 8 N.

To investigate the cause of the decreased mass estimate at low force levels in extension (Fig. 5), two subjects were fitted with splints which compressed the finger pad and prevented interphalangeal (IP) joint movement. As in the other experiments described above, eight trials were conducted at finger tip force levels of 1, 2, 4, and 12 N. Figure 7 shows the mass parameter estimate for one subject for these trials. The inertial contribution of the splint itself was measured and subtracted from the mass estimates computed in the splinted flinger trials. Instead of a rising mass estimate with force level at low forces as seen in the unplinted trials, we now see a nearly constant mass estimate at all force levels.

The damping ratio of every trial for all subjects for abduction was less than the corresponding damping ratio at the same force level for extension. Figure 8 shows the ratio of the mean ζ, for all subjects (ζ_{extension} / ζ_{abduction}) at each force level from 2 to 8 N. The mean ratio is consistently greater than 1.0 across this range.

IV Discussion

A Mass Estimate. Over the finger tip force range tested, the effective mass estimates for extension of each subject's finger either remained nearly constant or rose to a constant level of approximately 6 g. The diminished mass estimates seen at low finger tip force levels can be explained by a combination of two phenomena. First, at the lowest force levels, the subcun-

<table>
<thead>
<tr>
<th>Table 1</th>
<th>Subjects' mean and standard deviation (stdev) values of the parameters m, s, and ζ for three of six finger tip force levels in extension</th>
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<tbody>
<tr>
<td>Level</td>
<td>m (g)</td>
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<tr>
<td>2 N</td>
<td>3.8 ± 0.3</td>
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<tr>
<td>4 N</td>
<td>4.2 ± 0.4</td>
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<tr>
<td>8 N</td>
<td>4.8 ± 0.5</td>
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<table>
<thead>
<tr>
<th>Force</th>
<th>m (N/m)</th>
<th>s (N/m)</th>
<th>ζ</th>
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<tbody>
<tr>
<td>2 N</td>
<td>3.0 ± 0.2</td>
<td>2.5 ± 0.1</td>
<td>1.8 ± 0.1</td>
</tr>
<tr>
<td>4 N</td>
<td>3.5 ± 0.3</td>
<td>2.8 ± 0.2</td>
<td>1.9 ± 0.2</td>
</tr>
<tr>
<td>8 N</td>
<td>4.0 ± 0.4</td>
<td>3.2 ± 0.3</td>
<td>2.0 ± 0.3</td>
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Table 2 | Subjects' mean and standard deviation (stdev) values of the parameters m, s, and ζ for two of four finger tip force levels in abduction |
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<tr>
<td>Force</td>
<td>m (g)</td>
</tr>
<tr>
<td>2 N</td>
<td>3.8 ± 0.3</td>
</tr>
<tr>
<td>4 N</td>
<td>4.2 ± 0.4</td>
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of the joint beyond the horizontal orientation. At force levels greater than about 6 N, the finger stiffness is large enough to prevent this, and the finger behaves like a rigid body. For abduc-
tion, the estimated margin is not reduced, because the fingerpad along the medial edge of the index finger tip is not very compli-
ant, and because the IP joints do not move in the abduction–
adduction direction.

The extension trials carried out with two subjects with wooden splints verified that the low mass estimates were a result of the high finger tip pad and IP joint compliances at low force levels. The addition of the splint sufficed the effects of pad and IP joint compliance, which otherwise reduce the effective mass of the finger system.

B Damping Ratio Estimate. Because the effective mass is diminished at force levels below 6 N, the identified damping ratio for the extension trials is much larger than at high force levels. In addition, the ζ from these fast transient finger exten-
sion experiments is in notable contrast to previous single joint studies, which reported a strongly underdamped response with a damping ratio in the range 0.1–0.4 (e.g., ankle flexion–exten-
sion (Hunter and Kramers, 1982), and long-term finger abduc-
tion–adduction (Becker and Mote, 1990)). A factor that may contribute to the greater damping ratio identified in this study is the preclusion of reflex responses, due to the brief duration of the trial. Although the action of these reflexes varies with task parameters, in general, they substantially increase joint stiffness, particularly the stretch reflex (e.g., Akazawa et al., 1983; Doemens and Raad, 1993). Since the damping ratio var-
ies inversely with the square root of the stiffness, the absence of these reflexes results in a higher damping ratio. This suggests that across longer time scales, the damping ratio identified for the index finger would decrease, approaching values identified for other limbs. Further studies across larger time scales will be required to determine the magnitude of the stiffness contribution from the stretch reflex in this configuration.

Several biomechanical factors may influence the difference in the relative magnitude of the damping ratio identified for extension and for abduction as seen in Fig. 8. The unique physi-
ology of the fingers acting in flexion–extension may increase the damping ratio of the fingers in extension. Many of the previously analyzed joints, such as the elbow, are served by muscles adjacent to the joint with relatively short tendons. Like-
wise, abduction–adduction of the MCP joint is actuated by the intrinsics muscles that are intrinsic to the hand and have correspondingly short tendons. In contrast, flexion–extension of the MCP joint is actuated principally by the flexors digitorum superficialis and profundus and the extensor digitorum and indicis, located in the forearm. The tendons through which these muscles act pass through the carpal tunnel and palm, which may produce additional passive damping. Another factor for the large identified damping ratio may be the compliance of the tendons themselves. At the force levels in this study, tendon stiffness may make a significant contribution to the identified finger tip stiffness (Amis, 1994). This would add another de-
gree of freedom between the finger tip and the muscle. The finger tip pad can also be expected to provide additional damp-
ing at the contact of the apparatus, although this effect should be small above about 2 N. A more elaborate model would be required to quantify the contributions of these biomechanical factors; however, the high VAF values in this study nevertheless indicate that the second-order lumped element model provides a reasonable description of finger tip interactions for fast trans-
ients.

C Haptic Interface Design. For haptic interface applica-
tions, the plots showing variation of impedance parameters (Figs. 5 and 6) summarize the results. Of particular note in this context are: the near-constant value of the mass for single sub-
jects (in the >6 N range) and small variation between subjects; the approximately linear increase in stiffness with mean force level; and the relatively large and nearly critically damped value of the damping ratio for fast transients. One important aspect of these measurements for design purposes is that they represent likely minimum values for stiffness that will be encountered in task execution. As noted above, stretch reflexes probably act to increase stiffness, as do agonist–antagonist co-contractions. Thus, at longer time scales where reflexes are significant, these measurements probably represent lower bounds to the effective stiffness.

Mechanical impedance at short time scales may be particu-
larly important for multifinger haptic interfaces, in contrast to arm-level devices where frequencies above perhaps 2 Hz are of relatively less importance. Results presented by Howe and Kramers (1992) suggest that force feedback bandwidths of at least 8 Hz are helpful in teleoperated precision grasp tasks.

In such high-bandwidth systems, hand surface contacts or im-
pa
ges cannot produce rapid increases in stiffness. The mech-

Fig. 7 Comparison of mass estimate for splinted versus unsplinted ex-
tension trials for one subject. Means and standard deviations for each set of eight trials at each force level are depicted.

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Fig. 8 Normalized damping ratio (ζ = ζ,body) across force range shared by extension and abduction. Note that normalized ratio is always >1.
The combination of both finger and IP joint compliances, in varying degrees for different subjects, results in nonrigid body motion of the index finger and accounts for the reduced effective mass at lower finger tip force levels. This result dem- onstrates that it is not sufficient for successful identification of teleoperator inertia to approximate the effective inertia of a finger by measuring its mass, because at low force levels, the effective mass loading the manipulator may indeed be less than the entire finger mass.

The results presented here identify impedance variation with a simple parameter, tip force frequency, and specify quantitative values for the design of dexterous interfaces for teleoperation. Impedance will also vary with many other parameters, including joint angle, joint speed, co-contraction of antagonist muscles, input magnitude, and the time scale of the transient (Kanevon and Hunter, 1950). For fingers in particular, the impedance can be expected to vary with the angles of other joints in the wrist or in the kinematics of the fingers themselves. In addition, a complete representation of finger impedance will require measurements of responses at longer time scales.

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