# Dynamic Lumped Element Response of the Human Fingerpad

D. T. V. Pawluk[1]
e-mail: dianne@bme.jhu.edu

R. D. Howe

Harvard University,
Division of Engineering and
Applied Sciences,
Pierce Hall, 29 Oxford Street,
Cambridge, MA 02138

*The dynamic response of the fingerpad plays an important role in the tactile sensory response and precision manipulation, as well as in ergonomic design. This paper investigates the dynamic lumped element response of the human fingerpad in vivo to a compressive load. A flat probe indented the fingerpad at a constant velocity, then held a constant position. The resulting force (0–2 N) increased rapidly with indentation, then relaxed during the hold phase. A quasilinear viscoelastic model successfully explained the experimental data. The instantaneous elastic response increased exponentially with position, and the reduced relaxation function included three decaying exponentials (with time constants of approximately 4 ms, 70 ms, and 1.4 s) plus a constant. The model was confirmed with data from sinusoidal displacement trajectories.*

## 1 Introduction

The fleshy pad at the tip of the human finger mediates many of our mechanical interactions with the world. Because it acts as a coupling element between the hand and grasped objects, a complete explanation of precision manipulation must include the role of fingerpad deformation. Fingerpad mechanics are also a major factor in the tactile sensory response, which is an essential component of dexterity [7, 14]. Dynamic properties are particularly important in this context, as several of the specialized mechanoreceptors in the fingertip respond only to changing mechanical stimuli [7]. Fingerpad dynamics are also important in practical applications like the ergonomic design of vibrating tools [8] and in the analysis of repetitive tasks like keyboard typing [11, 12].

For ergonomic and rehabilitation purposes, a few researchers [8, 13] have measured the mechanical impedance of the fingerpad as a function of vibration frequency (20 Hz to 10 kHz) over a variety of preload forces (0.5 to 5.6 N). Similarly motivated, Serina et al. [11] examined fingerpad displacement as subjects applied low-frequency force sinusoids (0.25 to 3 Hz) to a flat plate at a contact angle of 45 deg. Gulati [4] and Gulati and Srinivasan [5] have more extensively measured the lumped response of the fingerpad to variously shaped indentors (point, cylinder, and flat plate) applied in position ramp and hold (0.5–32 mm/s) and sinusoidal (0.125–16 Hz) trajectories. In addition, Gulati proposed separate models of the lumped fingerpad for the different indentor shapes, each consisting of a series of five linear Kelvin models acting at different depths.

The main objective of this paper is to develop a model that describes the dynamic response of the fingerpad to arbitrary inputs within the range of 0 and 2 N applied by a flat plate. We chose a well-established quasi-linear viscoelastic model [2] as it had the potential to ascribe some meaning to both the functional form of the model and the model parameters. Our goal is to provide an understanding of the mechanical characteristics of the fingerpad that can be further used in understanding more complex systems in which the fingerpad is an important part. This includes both its use as a basis for understanding the mechanical component of human tactile sensing, and its importance as a component of a whole finger when studying grasping and manipulation. By confirming the model with a different type of displacement input (i.e., sinusoidal inputs), we verify its utility as a tool for predicting the response of the fingerpad to new types of displacement inputs, some of which may be more appropriate for these more complex analyses.

## 2 Methods

A motorized indentor applied controlled displacements to the fingerpad of the index finger. The response of the fingerpad to dynamic forces between 0 and 2 N was considered. Forces in this range are the most relevant for examining the response of the fingerpad during grasping [14] and are also the typical forces in typing [12]. Experiment 1 examined the force response of the fingerpad to constant velocity indentations. The results of this experiment were used to determine the appropriate velocity to use for system identification (i.e., the velocity above which rate-dependent effects were relatively small). Experiment 2 contained two different types of input trajectories. We first used a fast velocity ramp followed by a constant position hold phase for system identification. Then, in order to verify the model using a different type of input trajectory, we applied various sinusoidal displacement inputs to the fingerpad. A quasi-linear viscoelastic model is introduced to describe the results.

**(A) Experimental Apparatus.** The subject's outstretched hand was supported, palm upward, in a plastic mold lying horizontally on a table. Individual molds were constructed to allow subjects to rest their hands comfortably (which resulted in a variable amount of inward curl of each subject's hand and fingers). The resulting angle between the dorsum of the distal part of the index finger and the tabletop varied between subjects from approximately 20 to 40 deg. This configuration is similar to that used in many manipulation operations, as can be seen, for example, in grasping a flat-sided box. To preclude fingerpad movement, the plastic mold was closely fitted to the dorsum of the hand, the fingernail of the index finger was glued to the mold, and the hand and forearm were constrained using athletic tape.

The motorized indentor (see Fig. 1) applied controlled displacements vertically (with reference to the tabletop) to the fingerpad of the index finger with a flat probe. A two-axis strain gauge based force sensor (rms noise ≈ 3 mN) measured the forces applied vertically to the fingerpad and in one horizontal direction; the horizontal force signal served only to confirm the absence of significant shear forces. The signals were filtered with a two-pole analog low-pass filter with a 1 kHz cutoff

[1] Current address: Johns Hopkins University, Baltimore, MD.

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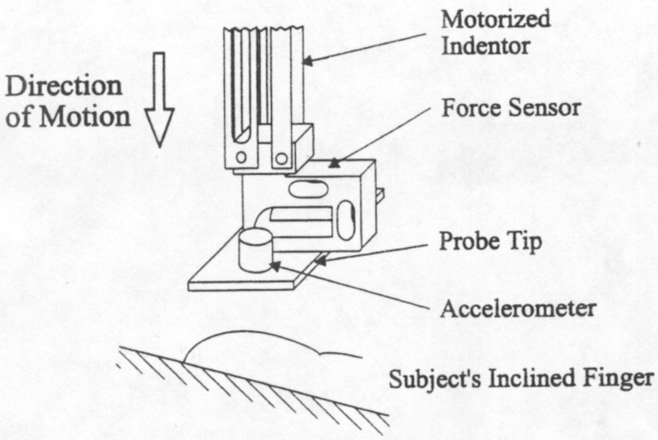


**Fig. 1 Side view of the tip of the experimental apparatus. A force sensor at the tip measured the force in the vertical and one horizontal direction. An accelerometer mounted on the tip measured the acceleration in the vertical direction. A joint sensor on the motor shaft measured the position.**

frequency. A magneto-resistive position sensor located on the motor shaft measured the position of the indentor (rms noise $\approx$ 3 $\mu$m). A 4.5 gram, low impedance piezoelectric accelerometer measured the acceleration of the indentor tip (rms noise $\approx$ 90 mm/s$^2$, resonant frequency = 22 kHz). The data were recorded using a 16-bit analog-to-digital converter at a sampling rate of 10 kHz; the quantization levels were below the rms noise of the sensors.

### (B) Experimental Procedure

*Experiment 1: Time Scale Determination: constant velocities.* Trajectories of constant velocity (0.2, 8, 16, 32, 48, 64, and 80 mm/s) were applied to the fingerpad to a force level of 2–3 N. The velocities were presented as a partial Latin square for a total of 28 trials (4 at each level). This order was: 0.2, 32, 64, 16, 48, 80, 8, 64, 16, 32, 8, 0.2, 48, 80, 32, 8, 80, 48, 64, 16, 0.2, 48, 0.2, 16, 32, 80, 8, 64. Six healthy subjects (1 female, 5 male; ages 24–36) voluntarily participated in this study. Slower speeds (0.1, 0.2, 0.5, and 1 mm/s) were also examined for two subjects (1 female, 1 male).

*Experiment 2: System Identification and Model Verification*

*(a) Identification: ramp-and-hold.* The fingerpad response was identified by applying a fast ramp at 60 mm/s to approximately 2–3 N, followed by a 5–7 s hold phase at the endpoint position. The objective was to examine time-independent effects with the fast ramp, and time-dependent effects with the hold phase [2]. 60 mm/s was chosen for the ramp as only small differences were found above this speed in Experiment 1. The duration of the hold phase was chosen by determining when the measured force essentially stopped decreasing. This point was defined as the time at which the force response decreased by less than 0.015 N in the previous half second; this force change corresponded to the peak-to-peak noise in the force response in holding a constant force level under active servo. Variability due to a subject's blood pressure variation was minimized by comparing points at similar phases of the pulse cycle.

*(b) Verification: sinusoids.* The system response to more general, but easily interpretable, inputs was investigated using sinusoidal trajectories. These trajectories were a 20 mm/s position ramp to a set operating point, followed by five cycles of either a 2, 4, 8, or 16 Hz position sinusoid, which spanned a force range from close to 0 N to approximately 2 N.

Four trials of the fast ramp-and-hold were followed by a Latin squares presentation of four trials each of the four different types of sinusoidal trajectories (16 trials), and then four trials, again, of the fast ramp-and-hold. The latter four trials were intended to verify the repeatability of the measurement across the duration of the experimental period. Four healthy subjects (1 female, 3 male; ages 20–28) voluntarily participated.

*General Considerations.* During all experiments, the fingerpad was allowed to recover from any viscoelastic effects for a minimum of 14 seconds between each trial. This was verified by determining the repeatability among the four trials of the fast ramp and hold experiment: Repeatability of the peak force response was within 3.5 percent and there were no ordering effects ($\eta$ = 0.17). However, the effective mass of the probe tip in front of the force sensor did affect the response significantly at high accelerations. In the case of the fast ramp and hold, it was necessary to adjust the force response to compensate for inertial effects due to deceleration. This was performed by measuring the acceleration of the probe tip and using it to calculate a correction for the measured force due to the inertial forces. The peak magnitude of the correction was approximately 15 percent. For all experiments, coupling of the force to directions perpendicular to the indentation was below 4 percent.

### (C) Model

*Model Description.* The model we propose to describe the force response is based on Fung's quasi-linear viscoelastic model of tissue [2]. It consists of two components: (1) an instantaneous elastic response, $T^{(e)}(x)$, which is the instantaneous force response of the fingerpad to a position step, $x$, from the undeformed position (which occurs so quickly that viscous effects do not have time to act); and (2) the reduced relaxation function, $G(t)$, which is the normalized, time-varying response of the fingerpad following the position step. The portion of the force response, $P_i(t)$, to an infinitesimal change in position, $\partial x$, at position, $x$, imposed at an instant of time, $\tau$, is described by

$$P_i(t) = G(t - \tau)\frac{\partial T^{(e)}[x(\tau)]}{\partial x}\,\partial x(\tau) \tag{1}$$

for $t > \tau$. We further assume that the resulting force response, $P(t)$, to an arbitrary input trajectory is the integral of the responses, $P_i(t)$, to a series of small position changes describing the input trajectory (i.e., superposition holds). Our system identification protocol, consisting of a fast position ramp followed by a position hold phase, was used to identify the two components of the model separately.

For many tissues, the elastic response, $T^{(e)}(x)$, can be modeled as an exponential function of position

$$T^{(e)}(x) = \frac{b}{m}\,[e^{m(x-x_0)} - 1] \tag{2}$$

where $b$ and $m$ are constants to be determined from the experimental measurements, and $x_0$ is the initial point for which $T^{(e)}(x)$ is experimentally observed to be zero [2]. We have also observed that the lumped force response of the fingerpad can be described by this exponential function (see Section 3).

Fung suggests that in general the reduced relaxation function includes an infinite number of time constants [2]. In practice this function can often be approximated using a finite number of terms. $G(t)$ is then represented by

$$G(t) = \frac{c_0 + \sum_{i=1}^{n} c_i e^{-\upsilon_i t}}{\sum_{i=0}^{n} c_i} \tag{3}$$

where $c_i$ are the proportion each term contributes to the force relaxation response and $\upsilon_i$ are the time constants.

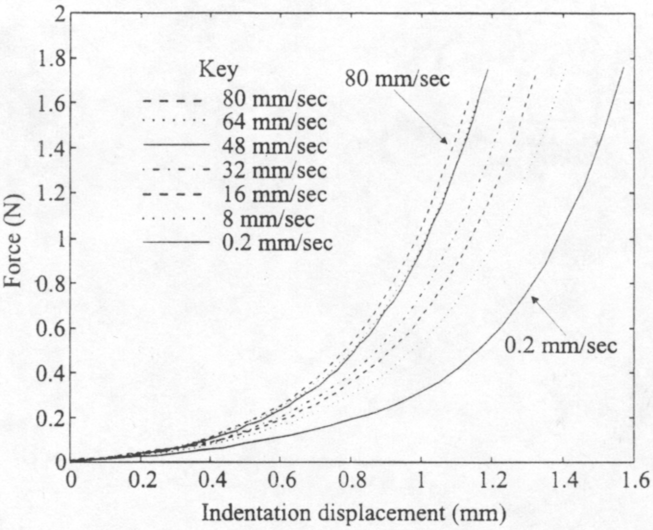


**Fig. 2 Force response of the fingerpad as a function of indentation depth of a typical subject to indentations of constant velocities. Data at each speed are from four trials, which were geometrically averaged. The average standard deviation of the four trials from the geometrically averaged response, across subjects and velocities, was 2.4 $\mu$m for position and 0.038 N for force.**

The resulting force response, $P(t)$, to an arbitrary applied displacement trajectory, $x(t)$, is then described by the convolution

$$P(t) = \int_{-\infty}^{t} G(t-\tau) \frac{\partial T^{(e)}[x(\tau)]}{\partial x} \frac{\partial x(\tau)}{\partial \tau} d\tau. \qquad (4)$$

*Quantification Method.* We can quantify the two components of the model by fitting the results of Experiment 2. As there is relatively little change in the force response at speeds above the one chosen for the fast ramp (see Section 3 and the observed response to the constant velocity indentations, Fig. 2), the fast ramp is a good approximation to the instantaneous response, $T^{(e)}(x)$, and can be used to characterize it. Because the two components are linearly combined in the model, the reduced relaxation function, $G(t)$, can be parameterized by the hold phase at a single position.

In more detail, the instantaneous response, $T^{(e)}(x)$, could be determined by obtaining a least-squares fit of the data from Eq. (2) for the parameters $m$, $b$, and $x_0$. However, this form of the response is difficult to parameterize due to the near-zero slope of the response at low forces (Fig. 2). Determining $x_0$ directly from the experimental records is also problematic for this reason and the inevitable presence of noise in the sensor signals.

These issues can be avoided if we differentiate Eq. (2) with respect to position and obtain stiffness as a function of force:

$$k = \frac{dT^{(e)}}{dx} = mT^{(e)} + b. \qquad (5)$$

This form is more reliable to parameterize as it is insensitive to the value of $x_0$, and requires a simple linear fit to $m$ and $b$. The stiffness is calculated from the experimental data by first smoothing the position and force trajectories with a nine-point symmetric moving average (0.9 ms width). Then the force is numerically differentiated with respect to position using a five-point interpolating polynomial for unequally spaced points [1]. Four trials are fit to Eq. (2) using a least-squares method.

The force response to the hold phase is fit to Eq. (3) to determine the parameters $c_0$, $c_i$ and $v_i$, $i = 1$ to $n$, of the reduced relaxation function, $G(t)$. Four trials are fit by a simplex method using MATLAB (The Mathworks, Natick, MA).

## 3 Results

### (A) Experimental Results

*Experiment 1: Time Scale Determination: constant velocities.* Data from the response of a typical fingerpad to indentations of constant velocities are shown in Fig. 2. In general, the force increases exponentially for all indentation velocities, with the slope increasing more rapidly with increasing speed. The response saturates at speeds above approximately 60 mm/s and below approximately 0.2 mm/s (slower speeds not shown).

*Experiment 2(a): Identification: ramp-and-hold.* The experimental data from all subjects for the system identification protocol showed an exponentially increasing force response to the fast position ramp, similar to the faster speeds in Fig. 2. The average standard deviation of the four repeated trials from the geometrically averaged response across subjects was 4.9 $\mu$m for position and 0.018 N for force. Plotting the calculated stiffness as a function of force (Fig. 3) shows that the stiffness increases linearly with force, in accord with Eq. (5).

For the position hold phase, subjects showed an exponentially decaying force response that approached a nonzero steady-state value within 5–7 seconds. During this latter phase, the effects of subjects' blood pressure variations at the pulse frequency (approximately 1 Hz) were clearly seen. A small artifact, due to the rapid deceleration of the indentor at the end of the ramp, is visible near 0.01 s. The average standard deviation of the force response within repeated trials across subjects was 0.026 N. Plotting the natural logarithm of the force response of the fingerpad during the hold phase as a function of time (Fig. 4) suggests that the reduced relaxation function, $G(t)$, can be reasonably approximated as a sum of three dominant time constants: one below 10 ms, another between 10–100 ms, and a third much slower term. In addition, a nonzero steady-state term is clearly required.

*Experiment 2(b): Verification: sinusoids.* The responses to sinusoidal displacement trajectories of differing frequencies exhibited the same general characteristics as the previous data. A response to a 4 Hz sinusoid is shown in Fig. 5. The increasing stiffness of the fingerpad with increasing indentation is visible in the distorted force response within each cycle: The positive peaks appear more "peaked" and the negative peaks appear more "rounded" than a sine wave of the corresponding fre-

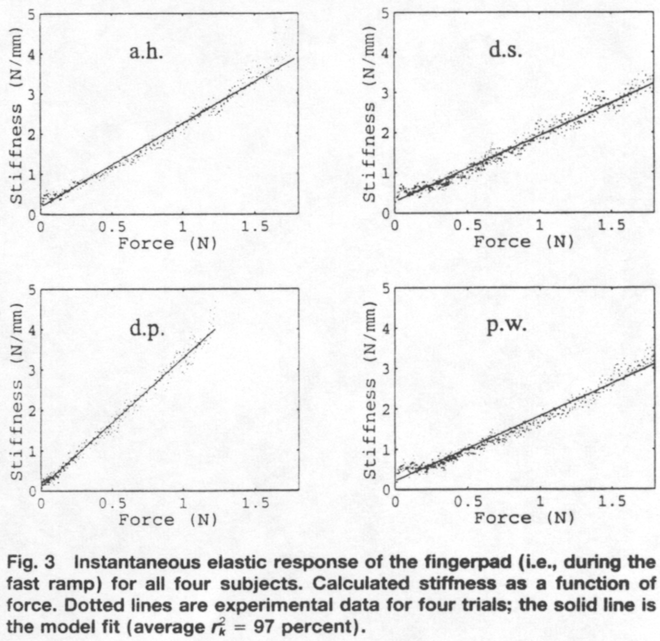


**Fig. 3 Instantaneous elastic response of the fingerpad (i.e., during the fast ramp) for all four subjects. Calculated stiffness as a function of force. Dotted lines are experimental data for four trials; the solid line is the model fit (average $r_k^2$ = 97 percent).**

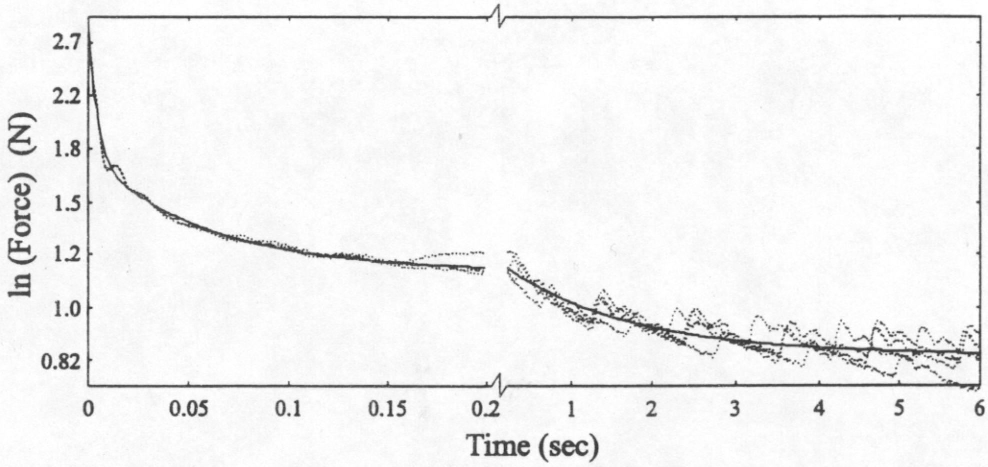


**Fig. 4 Force relaxation of the fingerpad of subject p.w. Dotted lines are four experimental trials; the solid line is the model fit ($r^2_{G(t)}$ = 89 percent). Blood pressure variations at the pulse frequency are clearly visible.**

quency. The relaxation of the force over time is, in part, discernible by comparing the peaks between cycles. The average standard deviation of the force response within repeated trials across subjects was 0.055 N; the variation in the peaks was approximately 12 percent.

**(B) Model Fit.** The two components of the proposed model, the instantaneous response and the force relaxation, were calculated from the results of Experiment 2 as described in Section 2(C).

*Calculation of the Instantaneous Response, $T^{(e)}(x)$.* The resulting linear least-squares fit of Eq. (5) to the responses for all subjects are shown in Fig. 3; the corresponding parameters are given in Table 1. The variation of the force response accounted for by the model, $r_k^2$, was very significant (on average, 97 percent). However, the large variation of the parameters $m$ and $b$ across subjects ($\approx$2:1) for this small sample suggests that we cannot make *quantitative* generalizations to the general population.

*Calculation of the Force Relaxation, $G(t)$.* The force response to the hold phase was fit to Eq. (3) to determine the parameters $c_0$, $c_i$, and $\nu_i$, $i = 1-3$, of the reduced relaxation function, $G(t)$. The resulting least-squares fits for all subjects

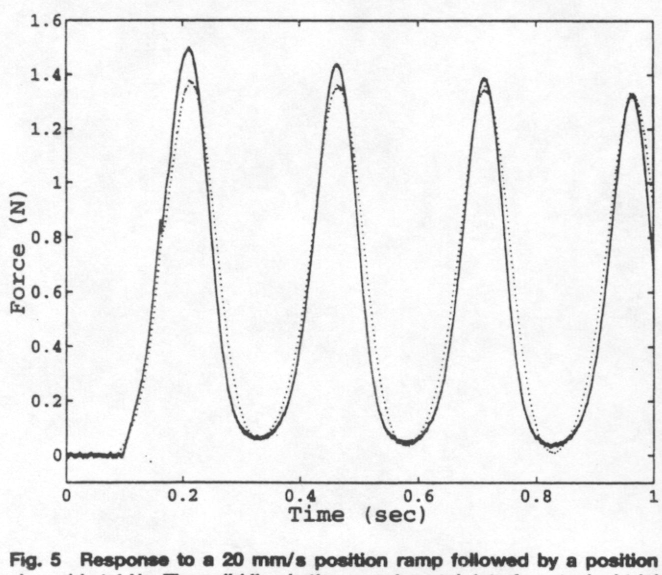


**Fig. 5 Response to a 20 mm/s position ramp followed by a position sinusoid at 4 Hz. The solid line is the experimental data for a typical trial and the dotted line is the model prediction.**

are given in Table 1 and shown for subject p.w. in Fig. 4. The variation of the force relaxation accounted for by the model ($r^2_{G(t)}$) was, on average, 95 percent. However, this somewhat underestimates the model's accuracy. Note that although the parameter fit for this subject had the lowest variance accounted for ($r^2_{G(t)} = 0.89$), the fit appears to be much better; most of the remaining variance is probably due to blood pressure variations at the pulse frequency. These variations are clearly visible in Fig. 4, and were present for other subjects as well.

It would be useful to correlate the fitted parameters for both $T^{(e)}(x)$ and $G(t)$ with easily measured variables, such as the width and thickness of the fingerpad. This might permit prediction of the fingerpad dynamics without the need to perform biomechanical measurements. In Table 1 we include the measured thickness ($t$) and width ($w$) of the finger at approximately the middle of the contact area. The slope of the instantaneous response, $m$, was negatively correlated with both the thickness and width of the fingerpad ($r^2_{mt} = 0.66$, $r^2_{mw} = 0.62$). This suggests that thicker and wider fingers are also softer. Correlations with other model parameters were less clearly interpretable due to their smaller variation between subjects. In addition, due to the small sample size, it is difficult to draw conclusions for the general population.

**(C) Model Confirmation.** The model was verified using the force responses to the sinusoidal position trajectories of Experiment 2. These data were not used to parameterize the model and their form is distinct from the ramp-and-hold trajectory, although they are sufficiently simple that interpretation is straightforward. The predicted force output, $P(t)$, was calculated from the measured position input, $x(t)$, using Eqs. (2), (3), and (4), with the parameters determined from the fast ramp and hold experiments (Table 1).

In addition to the parameters that had been previously determined, the initial starting point of the response, $x_0$, was needed. As it was difficult to estimate $x_0$ directly from the experimental results (see above), it was obtained by performing a least-squares fit of Eq. (2) to the fast ramp data; $x_0$ was the fitted parameter, with the values of $m$ and $b$ given in Table 1 used as constants. Using the resulting values of $x_0$ for the calculated predictions of the force response to the sinusoidal inputs produced a good fit to the data. The model response was calculated using the actual experimental displacement input applied and then filtered using a two-pole digital filter with a cut-off frequency of 1 kHz to replicate the effects of the analog filter on the force sensor. The average mean squared error (mse) was 7 percent. The maximum error typically occurred at the peaks of the sinusoidal inputs and was, on average, 18 percent. Although

**Table 1** Model parameters for individual subjects. The first two parameters describe geometric characteristics of the fingerpad: $w$ = width and $t$ = thickness. The second two parameters are from the fit of Eq. (2) to the instantaneous response data; V.A.F.$_{x}$ is the goodness of fit. The remaining parameters are the weightings and time constants from the fit of Eq. (3) to the force relaxation data; V.A.F.$_{\sigma(t)}$ is the goodness of fit.

| Subject | w (mm) | t (mm) | m (mm$^{-1}$) | b (N/mm) | V.A.F.$_{x}$ | $c_\infty$ | $c_1$ | $c_2$ | $c_3$ | $v_1$ (sec$^{-1}$) | $v_2$ (sec$^{-1}$) | $v_3$ (sec$^{-1}$) | V.A.F.$_{\sigma(t)}$ |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| d.p. | 12 | 9.5 | 3.2 | 0.092 | 0.98 | 0.22 | 0.45 | 0.15 | 0.18 | 253 | 14 | 0.66 | 0.99 |
| d.s. | 14 | 11.5 | 1.6 | 0.28 | 0.97 | 0.29 | 0.4 | 0.19 | 0.13 | 183 | 11 | 0.77 | 0.94 |
| a.h. | 14 | 10 | 2.1 | 0.18 | 0.98 | 0.25 | 0.4 | 0.19 | 0.16 | 224 | 14 | 0.63 | 0.96 |
| p.w. | 17 | 13 | 1.6 | 0.22 | 0.96 | 0.29 | 0.38 | 0.20 | 0.13 | 264 | 22 | 0.69 | 0.89 |
| means | 14 | 11 | 2.1 | 0.19 | 0.97 | 0.26 | 0.41 | 0.18 | 0.15 | 231 | 15 | 0.69 | 0.95 |

the error was larger at the peaks, it was comparable to the variation in the peaks between repeated trials in the actual experimental results, which was approximately 12 percent.

These model predictions can be improved by taking into account the small variations in the contact point between trials (presumably due to small movements of the fingertip within the mold, see Section 4). Due to the uncertainty in estimating $x_0$ directly from the results, the variability in the position at a measured force of 0.05 N was examined. The average variation in position across subjects was 80 $\mu$m. As this variation was significantly greater than the servo noise (which was typically less than 10 $\mu$m peak to peak), it is assumed that the variations at 0.05 N are correlated to those at the contact point. The value for $x_0$ used in predicting the force response was therefore allowed to vary in the fitting algorithm within a range of 80 $\mu$m between trials for each subject. In addition, variations were correlated in size and direction with the positional change determined at 0.05 N. The average mse of the new predictions improved to 4.5 percent; the error at the peaks decreased to 10 percent. A representative comparison between the predicted model output and the experimental data is given in Fig. 5; the mean squared error (mse) was 4 percent.

## 4 Discussion

**Interpretation of the Experimental Results and Model Choice.** In this study, we first examined the force response of the fingerpad to displacement indentation trajectories of constant velocity. In general, the force was found to increase exponentially with position for all indentation velocities; the slope increased more rapidly with increasing speed. The saturation of the response at speeds above approximately 50 mm/s corresponded to the instantaneous stiffness of the fingerpad, before the time-dependent components had a significant effect. The saturation of the response below speeds of approximately 0.2 mm/s corresponded to the steady-state stiffness, after the time-dependent components had largely died out. In between, time-dependent effects made a substantial contribution to the force response.

The existence of a static stiffness suggests that even the simplest model describing these results must contain a spring. In the most parsimonious case, its stiffness must be exponential as a function of displacement to describe the quasi-static force response correctly. As there are also changes in the force response with changes in velocity, the model must also include a damper. For there to be a static response, the damper must be in parallel with the spring. Again in the most parsimonious case, the damping constant must be exponential as a function of displacement to maintain the exponential force response across all constant velocity trajectories. However, this model is incomplete as it predicts that the force response increases without bound with increasing velocity. It is therefore necessary to add a spring in series with the damper, which limits the maximum force response as a function of velocity. For the instantaneous stiffness to also be exponential with position, the spring constant of this series spring also needs to be exponential as a function of position. As a result, the most parsimonious model that can conceptually describe the force response to the set of trajectories of constant velocity is a Kelvin model with components exponentially dependent on position.

These experimental observations are not limited to only constant velocity trajectories. The nonlinear dependence of the force response with position is also apparent in the force response of the fingerpad to sinusoidal trajectories. In the linear case, one would expect the force response to be sinusoidal with the deviations of the positive and negative peaks from the median being equal. Instead, the actual results show a distorted sinusoidal force response with the positive peaks appearing more "peaked" and the negative peaks appearing more "rounded." In addition, the deviations of the positive peaks from the median are greater than that of the negative peaks.

The hold phase of the fast-ramp-and-hold trajectory can provide a picture of the time-dependent effects. Our examination of the hold phase suggested that there are several, rather than one, dominant time constants. In order to include this observation, we can extend our simple model to contain a series of Kelvin models, each with exponential components. This is in fact a structural interpretation of Fung's model for tissues of constant cross-sectional area [2]. In using Fung's model, we further assume that these time constants are linear.

**Model: Effectiveness of Fit, Interpretation, and Limitations.** The model we have presented works well in predicting the functional form of the lumped force response of the fingerpad. We have also found it to be effective (average mse = 7 percent) in quantitatively predicting the response of fingerpads to trajectories not used for model parameterization (i.e., sinusoidal displacement inputs). These model predictions can be improved upon by taking into account the small variations in the contact point between trials (average mse = 4.5 percent). The largest error in the prediction occurred during the first two cycles of the response. We believe that this inaccuracy is due to small errors in fitting the force relaxation function. To calculate the resulting force response in our simulations we take the instantaneous response (a large quantity) and decrease it by an amount depending on the past history (also a large quantity), producing a relatively small result. In doing so, small errors in the force relaxation function get magnified in the final result.

The efficacy of Fung's quasi-linear viscoelastic model is noteworthy, given that it was developed for tissues of constant cross-sectional areas in tension. Here we considered a biomaterial of changing cross-sectional area in compression. One possible explanation for the model's success for compression is that as the fingerpad is compressed, the collagen fibers may bow out and stretch as in tissues under tension. The fact that this model was effective for a biomaterial of changing cross-sectional area was initially very surprising. However, in [9] we show why this result is plausible despite the significant change in the contact area with indentation amplitude: The dynamic contact distribution appears to combine the localized tissue responses to produce the lumped response in a manner that retains the functional form of the underlying tissue.

Given the success of our model for the subjects in our experiment, it would be desirable to be able to generalize our results quantitatively. However, although the estimated parameter values fell within a fairly narrow range, the small sample size makes it difficult to generalize parameter values to the entire population *quantitatively*. These parameters are also expected to be affected by several other variables. The elementary mechanical properties of the biomaterial constituting the fingerpad (such as the elasticity of the fibers and the fluid content of the tissue) change with hydration and temperature. Also, preconditioning of the fingerpad is expected to have an effect, although the results are expected to be similar (cf. our results to Gulati [4] who used preconditioning).

Structurally, the bone of the distal phalange is modified significantly toward the tip, enlarging into a disk at the end. In addition, the curvature of the finger becomes greater toward the distal end of the fingerpad. These structural variables are expected to play a role in the variation of the force responses for varying finger inclination angles, even for relatively small variations such as in our experiment. An increase in the finger inclination angle moves the contact area closer to the tip, thereby changing the underlying structure. Part of the resulting effect was observed anecdotally in one subject, for whom their hand mold was remade and the finger inclination angle changed by about 20 deg: increasing the finger inclination angle significantly increased the steepness of the force response as a function of displacement. Further work will be required to quantify this effect, although the present results have defined the essential force–motion relationship.

The parameters estimated are also effected by the finite time response of the motorized indentor, which inevitably limits the accuracy of producing the system identification protocol (i.e., a fast ramp-and-hold trajectory). The primary limitation was the finite time to decelerate and hold a constant position. This probably resulted in a small underestimation of the force relaxation function.

Additionally, valid model predictions are limited to trajectories that remain in contact with the fingerpad. Breaking contact during retraction produces erroneous results. However, the calculation of the model response to sinusoidal displacements showed that the model is successful in predicting both loading and *unloading* of the fingerpad even at low forces. The ability to predict the instant when contact is broken in dynamic interactions may be of use in designing human tactile sensing experiments [3].

**Human Motor Control Considerations.** The qualitative form of the experimental results, as elucidated by our model, shows several desirable properties for human motor control. At low forces, the fingerpad is very compliant. This facilitates initial grasping of an object by avoiding large forces that might disturb the object before its position is precisely determined. However, highly compliant fingers would make manipulation difficult due to the uncertainty of the object's position as measured by the joint receptors. This problem is passively alleviated by the increasing stiffness of the fingerpad as the grasp force is incremented. Furthermore, damping increases in proportion to the stiffness (because the reduced relaxation function is linear) thereby maintaining stability with increasing force.

In addition, high stiffness at high grip forces is useful for rejecting disturbances. The stiff coupling between the finger and object lessens relative displacements as forces are applied to the object in the course of a manipulation task. Preservation of this relationship is also aided by the relatively low stiffness of the joints of the finger [6]. The effective tip stiffness of the index finger in extension at 2 N is over an order of magnitude less than the fingerpad in compression (0.14 versus 4.4 kN/m) and increases more slowly (40 versus 2100 $m^{-1}$). Therefore, if forces are applied to the object, the fingertips remain in contact and the joints accommodate the disturbance, minimizing the change in contact force.

In general, our experimental results are consistent with those obtained by Gulati and Srinivasan [5] and Serina and colleagues [11], who have examined the response of the fingerpad in similar force and frequency ranges. The model we have developed is comparable in accuracy to that developed by Gulati [4]. However, our model builds upon well-developed biomechanical theory that has been widely validated for other biomaterials [15]. In this paper, we have further shown that our model is capable of quantitatively predicting the response of the fingerpad to displacement trajectories different from those used to parameterize the model.

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