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Viscoelastic Characterization of the Primate Finger Pad In Vivo by Microstep Indentation and Three-Dimensional Finite Element Models for Tactile Sensation Studies

*When we touch an object, surface loads imposed on the skin are transmitted to thousands of specialized nerve endings (mechanoreceptors) embedded within the skin. These mechanoreceptors transduce the mechanical signals imposed on them into a neural code of the incident stimuli, enabling us to feel the object. To understand the mechanisms of tactile sensation, it is critical to understand the relationship between the applied surface loads, mechanical state at the mechanoreceptor locations, and transduced neural codes. In this paper, we characterize the bulk viscoelastic properties of the primate finger pad and show its relationship to the dynamic firing rate of SA-I mechanoreceptors. Two three-dimensional (3D) finite element viscoelastic models, a homogeneous and a multilayer model, of the primate fingertip are developed and calibrated with data from a series of force responses to micro-indentation experiments on primate finger pads. We test these models for validation by simulating indentation with a line load and comparing surface deflection with data in the literature (Srinivasan, 1989, "Surface Deflection of Primate Fingertip Under Line Load," *J. Biomech.*, **22**(4), pp. 343–349). We show that a multilayer model with an elastic epidermis and viscoelastic core predicts both the spatial and temporal biomechanical response of the primate finger pad. Finally, to show the utility of the model, ramp and hold indentation with a flat plate is simulated. The multilayer model predicts the strain energy density at a mechanoreceptor location would decay at the same rate as the average dynamic firing rate of SA-I mechanoreceptors in response to flat plate indentation (previously observed by Srinivasan and LaMotte, 1991 "Encoding of Shape in the Responses of Cutaneous Mechanoreceptors," *Information Processing in the Somatosensory System (Wenner-Gren International Symposium Series)*, O. Franzen and J. Westman, eds., Macmillan Press, London, UK), suggesting that the rate of adaptation of SA-I mechanoreceptors is governed by the viscoelastic nature of its surrounding tissue. [DOI: 10.1115/1.4029985]*

1 Introduction

Tactile information in humans and primates is encoded in the response of thousands of peripheral nerve endings embedded in the skin. A load applied on the surface of the finger pad is transmitted to mechanoreceptor locations within the skin. The resulting mechanical state (stress/strain) around the mechanoreceptor causes it to "fire," generating a train of neural impulses sent to the central nervous system and enabling us to interpret touch. These neural impulses, generated by the spatial distribution of responding mechanoreceptors, are in the form of a temporal sequence of action potentials and tactile information is encoded in the frequency of the generated action potentials. The neural response of mechanoreceptors in humans and primates to a "step" stimulus (a step force or step displacement applied on the finger pad with a blunt probe) has been characterized in the literatures [1–5]. For SA-I mechanoreceptors (Sec. 4.1), there are two distinct phases of neural response to a step indentation on the finger pad: a

dynamic phase and a static steady-state phase. The dynamic phase is observed immediately following stimulation, where the firing frequency of mechanoreceptors is initially high followed by a gradual reduction. This firing frequency reduces and eventually settles into a steady-state value. Most models in literature describing the primate finger pad mechanoreceptor neural behavior [6–9] try to relate the incident loading and resulting mechanical state at the mechanoreceptor to the steady-state firing rate or average firing rate (in a chosen time window) of the mechanoreceptor. Recent studies include Gerling et al. [10], where a finite element model of a human distal phalange was developed along with a receptor model to predict the average firing rate in the dynamic phase and static phase and to match available primate empirical data [1]. The average firing rates in the dynamic phase were calculated based on the inverse of the average interspike intervals on the dynamic phase (30–50 ms which includes the highest firing frequency poststimulation), while the average static firing rate was based on spikes observed in a larger time window, 650–950 ms [10]. To the best of our knowledge, limited literature is available that characterizes the transient firing rate of the mechanoreceptor on stimulation during the dynamic phase (i.e., characterizing the change in firing rate with time in the dynamic

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Fig. 1 Block diagram of the indentation apparatus used for characterizing the viscoelastic properties of the primate finger pad. The inset image shows the Delrin probe tip used for the indentation experiments. The diameter of the flat end of the cylindrical indenter is 0.5 mm. A typical input displacement stimulus and observed force output (shown qualitatively, not to scale). The indenter was not glued to the primate finger pad during experiments; however, on retraction of the tip from the skin some adhesion was observed.

phase before steady-state). This paper characterizes the viscoelastic behavior of the skin tissue and studies its relationship to the dynamic firing rate of SA-1 mechanoreceptors.

To develop a quantitative understanding of how spatiotemporal loads imposed on the surface of the skin are transmitted to mechanoreceptor locations within the skin, it is imperative to fully understand and characterize the geometry as well as mechanical properties of skin and its underlying tissues. One of the interesting problems in tactile sensation is to identify what stress/strain state around the mechanoreceptor causes it to respond. To understand this, we must try to relate the mechanical state around the mechanoreceptor due to an imposed load to the neural impulses that result from the stimulation. The experimental challenges in obtaining mechanoreceptor responses to mechanical stimuli in humans warrant the use of other model organisms to study tactile sensation. The wealth of tactile neural data available in primates [1,2] along with the similarity of the structure of the primate fingertip to that of humans makes it a popular model to study touch in humans.

There has been considerable progress in the development of accurate elastic biomechanical models of the primate and human finger pad. Early models idealized the finger pad [3] as an incompressible, homogenous, isotropic linearly elastic half-space. The “waterbed model” [11] assumed the finger pad to be an elastic membrane under pressure and was successful in accurately predicting human and monkey finger pad surface deflection to line loads. However, this model could not explain the transduction of mechanical signals into neural codes. To answer this, two-dimensional [6,7,12] and three-dimensional finite element models [8,13] of the human and monkey fingertips with realistic external geometry and internal layered structure of the skin and subcutaneous tissues were developed, to gauge the role of skin biomechanics in tactile response. The models developed by Dandekar et al. [8] used a linear elastic model for the skin tissue where a Poisson’s ratio of 0.48 was used (considering the tissue to be almost incompressible). The elastic moduli of the different layers were obtained by matching numerical experiments with available empirical data. Using these models, the strain energy density at the mechanoreceptor locations was shown to be the likely strain measure encoded by the mechanoreceptors.

1.1 Viscoelastic Characterization of Primate Skin Tissue. The work cited in the previous section assumed the mechanical behavior of human and primate skin tissue to be linearly

elastic. However, skin is well known to be viscoelastic and anisotropic in nature. We are interested in determining the mechanical state at mechanoreceptor locations at the advent of mechanoreceptor stimulation taking into account the viscoelastic nature of skin tissue. The advent of mechanoreceptor stimulation occurs at small strains [2] and thus there is a need to accurately characterize viscoelastic behavior of skin tissue at these small deformation ranges. There has been progress in the development of viscoelastic models of the human fingertip to study its response to a variety of dynamic mechanical stimuli [14–17]. However, to the best of our knowledge there is limited literature on empirically validated viscoelastic models of the monkey fingertip.

Due to the difficulty in isolating biological tissue specimens along with the challenges in preserving mechanical integrity of tissues in vitro, it becomes necessary to characterize material properties using in vivo methods. In this paper, we present a method to estimate mechanistic viscoelastic parameters of primate skin tissue in vivo for small strains by studying the stress relaxation behavior of the finger pad of a primate using a combination of single point indentation experiments and numerical simulation. We characterize the viscoelastic behavior of the finger pad in response to precise microstep indentation using a calibrated system with position resolution of $1\text{ }\mu\text{m}$ and force resolution of 0.3 mN and study the variation of this behavior across different fingers of an anesthetized primate. To avoid noise in the data due to motion artifacts during the experiments, we studied the viscoelastic response of the finger pads of different fingers of an anesthetized primate. To check the variability of our observed data, we model the empirical force–time response using a Maxwell–Weichert element and compare the fitted model parameters. In order to determine more mechanistic viscoelastic parameters, we develop two 3D multilayer finite element models of the primate fingertip: (a) a homogeneous viscoelastic model where all the tissues are modeled with a two term Prony series and (b) a multilayer model where the epidermis is modeled using an incompressible (Poisson’s ratio = 0.48) linearly elastic material, and the dermis and inner tissues are modeled with a two term Prony series. We build these models with realistic geometry from previously published data [8] and use them to simulate two sets of indentation experiments: (a) indentation with a cylindrical indenter and (b) indentation with a line load. Both the models are calibrated by matching the force–time response curves of our simulation with our experimental data (step indentation with a cylindrical indenter). In order to validate the models, surface deflection

Fig. 2 Position calibration of the indenter. (a) The setup used to determine the position calibration constant. Note that the photo does not show the PDMS block that was on top of the micrometer positioning “Z” stage used for calibration. (b) The calibration plot of the measured voltage versus the displacement controlled by adjusting the Z stage.

profiles of the model to line loads are matched with data available in the literature [11]. Using these methods, we calibrate and compare a homogeneous viscoelastic model and a multilayer elastic viscoelastic model described in Sec. 4. Finally, we indent the calibrated and validated model with a flat plate and compare the strain energy density versus time with available neurophysiological data in the literature [2]. Results are presented in Sec. 4.1 and reveal that the rate of adaptation of slowly adapting (SA) mechanoreceptors may be linked to the viscoelastic relaxation of the surrounding skin tissue.

2 Methods

2.1 Experimental Setup. The indentation apparatus consisted of an Aurora Scientific 300B Dual-Mode Lever Arm System, a custom designed 3 cm lever arm with a 0.5 mm diameter flat-tipped cylindrical indenter, and a Pentium 4 computer (Windows XP) equipped with a digital acquisition card (National Instruments PCI-6120) and MATLAB [18,19]. A custom program was written in MATLAB to control the position of the indenter (motor), while measuring and recording the actual position and the force response at the indenter tip.

This was done by commanding the desired position set point voltage to the Aurora controller as a function of time, while monitoring the controller’s position and force outputs. The position output is based on the signal from a position sensor attached to the motor shaft, and the force output is derived from the motor current via an analog circuit which compensates for system dynamics. The apparatus has a maximum indenter displacement of 10 mm, with a position resolution of $1 \mu\text{m}$ and is capable of measuring and delivering a maximum force of 0.5 N with a force resolution of 0.3 mN. Note that the force offset was set to maximum, so that the force output was not limited by the controller. A schematic of the experimental setup is shown in Fig. 1.

2.2 Calibration. Tests were performed to measure position and force calibration constants for the indenter apparatus. In the position calibration test, we attached a micrometer positioning stage (Edmund Industrial Optics, NT53-856) to the apparatus (Fig. 2(a)) and observed the position input voltages required to move the indenter to known distances. Specifically, a MATLAB program was written in which the surface of an object placed on the stage (a polydimethylsiloxane (PDMS) block, not shown in the figure) was detected by moving the indenter toward the object until the force output reached a particular threshold value (0.3 V).

The same threshold value was used for all position calibration measurements. The PDMS block was moved from 0 to 800 μm in fixed increments (160 μm) and the voltage required to detect the position of the PDMS block was measured and plotted (Fig. 2(b)).

To determine the force calibration constant (force–voltage relationship), the indenter was inverted (Fig. 3(a)), and the force output voltage was measured for different weights attached to the indenter tip. A torque balance about the pivot O of the indenter (refer to Fig. 3(a)) gives us

$$Wl \sin \phi + mg \frac{l}{2} \sin \phi = T \quad (1)$$

where W is the applied weight, l is the length of the indenter, mg is the weight of the indenter, and T is the torque at the motor. During calibration and in all the experiments, the angle ϕ was $90 \pm 10 \text{ deg}$ ($\pm 10 \text{ deg}$ uncertainty in ϕ corresponds to an uncertainty of $\pm 1.5\%$ in our force measurements), so the above equation can be rewritten as

$$W + \frac{mg}{2} = \frac{T}{l} = (V + V_o)k \quad (2)$$

where V is the measured voltage, V_o is the measured voltage when no load is applied ($W = 0$) at the indenter, and k is the calibration constant for $\phi = 90 \text{ deg}$. The calibration plots are given in Figs. 2(b) and 3(b). The force and displacement calibration constants were determined to be 0.015 V/mN and 2.042 V/mm.

2.3 Indentation Experiments and Results. Indentation experiments were performed on each of the five fingers of an anesthetized primate (*rhesus macaque*). The animal used for the experiments was a pair-housed, intact male rhesus macaque, weighing 9.5 kg and approximately 7.5 yr of age. It was pre-anesthetized with atropine (0.04 mg/kg IM) and anesthetized with a combination of ketamine (10 mg/kg IM) and xylazine (0.5 mg/kg IM). All animal procedures were performed in accordance with the National Institutes of Health guidelines and the Massachusetts Institute of Technology Animal Care and Use Committee.

The indentation experiment consisted of five independent repeated trials at four different indentation depths for each finger (20 trials per finger). During the experiment, the anesthetized primate was on its back facing up with its arm naturally extended out laterally with the extended forearm resting on an adjacent side table (palm facing up) at the same level as the bed. The finger

Fig. 3 Force calibration of the indenter. (a) Presents a schematic of the inverted indenter tip used to derive the force calibration constant and (b) the calibration plot of the measured voltage versus the applied weight on the indenter.

being investigated was then placed on a block (on the table and aligned with the indenter apparatus). The nail of the finger being investigated was adhered to the block using two-sided tape (the same boundary condition used in our simulations). We avoided restraints on the arm or hand or fingers to avoid any undue stretches in the skin that could have influenced our results. Even though the primate was anesthetized, passive motion artifacts were observed in the data if the primate's arm was not in a relaxed posture during the trial. This was seen during the first set of trials in the experiment (test with the index finger) and as a result these data were discarded and only ten trials with the index finger were successfully recorded. Adjusting the forearm to lay in more natural posture resulted in mitigating these artifacts. In the case of the other four fingers, the first trial was discarded due to setup issues associated with making sure the primate's arm was suitably adjusted and each finger had 19 independent trials. During each trial, the force response of the finger pad was measured and tabulated.

Static indentations were performed at depths of 200 μm , 400 μm , 600 μm , and 800 μm . The first step in each trial was for the flat-ended cylindrical indenter (0.5 mm diameter) to detect the skin surface. To do this, the program lowered the indenter at a velocity of 1 mm/s toward the skin until a 10 mN force threshold (0.1 V) was detected indicating contact with the skin. From this starting position, we began indentation which consisted of a 2-s hold to ensure equilibrium followed by a step input, which comprises a steep indentation ramp (5 mm/s) into the skin followed by a hold (7 s) and then a retraction ramp until the indenter left contact with the finger pad. The data acquisition rate of the indenter was 8000 samples per second. Between two successive indentations, there was a 2–3 min delay to allow the finger pad tissue to relax from viscoelastic effects. No two indentations were performed at the same location but all the indentations were within a central region of the given finger pad. Figures 4(b)–4(f) show the force responses of the different finger pads to step indentations of different depths. Figure 4(a) shows the repeatability of observations for one such set of measurements, namely, 0.80 mm of static indentation on the little finger pad of the primate.

3 Modeling of Viscoelastic Response (Empirical)

We model the viscoelastic behavior of the finger pad using a variation of the Maxwell–Weichert element. The element consists of three elastic springs and two purely viscous dampers as shown in Fig. 5(a). In our tests, we imparted an input displacement (step input) to the primate finger pad and observed the force response. To derive the response of the element to the same stimuli, consider the model depicted in Fig. 5(a). We need to derive the

expression for the force response, $F(t)$, of the element when a step input, $X(t) = X_0 \times H(t)$, where $H(t)$ is the heaviside function ($H(t) = 1$ for $t > 0$ and 0 for $t \leq 0$), is imparted to the system. The total force, $F(t)$ for a given step input, is the sum of the force contributions of the three arms of the element

$$F(t) = F_1(t) + F_2(t) + F_3(t) \quad (3)$$

Arm 1 of the modified Maxwell–Weichert element is an elastic spring with constant E_0 , arms 2 and 3 are Maxwell elements with spring constants E_1 and E_2 and damper constants η_1 and η_2 .

We formulate the differential equations, and on solving we get

$$F(t) = E_0 X_0 + E_1 X_0 e^{-(E_1/\eta_1)t} + E_2 X_0 e^{-(E_2/\eta_2)t} \quad (4)$$

which can be rewritten as

$$F(t) = A_0 + A_1 e^{-(t/t_1)} + A_2 e^{-(t/t_2)} \\ = A_0 \left(1 + \frac{A_1}{A_0} e^{-(t/t_1)} + \frac{A_2}{A_0} e^{-(t/t_2)} \right) \quad (5)$$

where t_1 and t_2 are time constants. It is to be noted that the parameter A_0 also corresponds to the quasi static force response or the force response when all viscous effects die down. This parameter can be used to estimate the effective stiffness of the finger pad. Figure 5(b) shows one such fit of this model with experimental data. The bold curve shows the predicted force response of the ring finger pad when indented with a step indentation of 600 μm .

3.1 Data Variability: Model Fit Parameters for Different Fingers. In order to estimate the variation in our experimental results, each set of experimental data was curve fit with the Maxwell–Weichert model and parameters were extracted and tabulated. All curve fittings were done using the commercially available graphing software application, ORIGIN (by Origin Lab Corporation, Northampton, MA), which uses a chi-square minimization algorithm and Levenberg–Marquardt (L–M) algorithm to estimate parameter values via an iterative procedure [20]. Figure 5(b) shows one such fit using the software package with $R^2 = 0.99$. Similarly, curve fitting was done for all 86 independent force curves (obtained by our experiments across four different indentation depths and across five fingers) with similar quality of fit ($R^2 > 0.95$) to obtain model parameters. It can be seen in Fig. 6(a) that there is not much variation of the two decay constants t_1 and t_2 across the fingers compared to what is typically observed with in vivo biomechanical data. The mean decay constant t_1 was

Fig. 4 Force response of the primate to static step indentation. (a) Illustrates the repeatability of the measurements by superimposing five independent trials of the force response of the little finger to a static indentation of 0.8 mm. (b)–(f) The force responses of the thumb, index finger, middle finger, ring finger, and little finger in response to step inputs of 0.2 mm, 0.4 mm, 0.6 mm, and 0.8 mm.

found to be 2.279 ± 0.233 s and $t_2 = 0.149 \pm 0.022$ s. As shown in Fig. 6(b), there is not much variation of the two dimensionless constants (A_1/A_0 and A_2/A_0) across all the fingers. These parameters were determined to be $A_1/A_0 = 0.371 \pm 0.082$ and

$A_2/A_0 = 0.294 \pm 0.059$ for all fingers across all indentation depths. We also determined the quasi static force parameter A_0 for different indentation depths. A_0 was found to vary linearly with depth with mean stiffness of 0.120 mN/ μ m.

Fig. 5 Model fit. (a) A variation of the Maxwell–Weichert model comprises three elastic springs and two purely viscous dampers. (b) The model fit to the force response of the ring finger pad of the primate in response to a step input (displacement) of depth 0.6 mm.

Fig. 6 Estimation of model parameters. (a) The two exponential decay parameters for all five fingers and (b) the dimensionless coefficients. Each point represents a mean of $n = 19$ readings taken across all four indentation depths (except for the index finger which includes only 10 successful trials; see Sec. 2.3). All error bars are ± 1 standard deviation and the finger digits are defined as follows: (1) index finger, (2) middle finger, (3) ring finger, (4) little finger, and (5) thumb. The plots show that for our given indentation ranges (200–800 μm) there is not much finger–finger variation in the model parameters and there are two distinct time regimes. In the short term ($\sim < 3\text{ s}$), the viscoelastic behavior is like a viscous liquid with fast stress relaxation governed by the smaller time constant. In the longer term ($> 3\text{ s}$), unlike a viscous liquid with small time constant, the decay slows and the stress does not entirely reduce to zero instead slows toward an asymptotic value governed by the springs in the model.

4 Estimation of Mechanistic Viscoelastic Parameters: Development of a 3D Finite Element Model

Srinivasan and coworkers [8] developed a 3D model of the distal phalanx of the primate with realistic geometry using a video-microscopy setup and image reconstruction algorithms. A sequence of 2D boundary images of primate fingertip replicas was extracted at different angles and a 3D model was reconstructed from the 2D cross sections layer by layer using this data. Each cross section consisted of five layers where the innermost layer was the bone. The skin was modeled with two layers corresponding to the dermis and epidermis and its dimensions were chosen from available data on the gross thickness of the skin. The dimensions of the bone were extracted from published X-ray images. Two more layers were constructed between the bone and the skin to model the adipose tissue and fibrous matrix. Using these

models, it was found that a three-layer model was adequate to predict surface deflection profiles to line loads [11]. The ratio of the elastic modulus of the five layers in the model was $10^4:10^3:10^3:10^3:10^8$, the last layer being the bone.

Using this available geometric data, we rebuilt the multilayer model of the primate fingertip in ADINA ver. 8.7.2 (Fig. 7). The model was meshed using 10,820 eight-node solid brick elements and consisted of 283,441 nodes. A cross section of the model, showing the four tissue layers is shown in Fig. 7(a) (the innermost layer not shown in Fig. 7(a) is the bone and was modeled as rigid). We used this model to simulate two sets of independent indentation experiments; indentation with a cylindrical indenter (Sec. 2.3 of this paper) and indentation with a line load [11]. To simulate the indentation with a line load, a flat rigid indenter of width 0.02 mm and length 5 mm was modeled in ADINA (Fig. 8(a)). To check our finite element model development, we repeated the

Fig. 7 (a) Cross section of the multilayer viscoelastic model built in ADINA showing the four layers of tissue, labeled L1–L4, with L4 being the outermost layer. The innermost layer of the model, the bone, is not shown such that the finger appears hollow in the figure. For scale, the monkey fingertip width at the cross section is ~8 mm. (b) The full 3D model used for simulating the indentation experiments.

Fig. 8 Line load indentation. (a) The model simulating the line load indentation experiment reported by Srinivasan [11]. (b) The deformed mesh of the primate finger pad indented to 0.5 mm after 3 s. The legend shows displacement in millimeter.

simulations performed by Dandekar et al. [8] assuming the tissue to be linearly elastic with Poisson's ratio 0.49 for a line load and match our results with that in the paper. As was reported by Dandekar et al. [8], a tissue elasticity ratio of $10^4:10^3:10^3:10^3$ best matched the surface deflection data.

The mechanical behavior of skin tissue is considered to be non-linear and viscoelastic [11,21]. The purpose of the present study is to develop realistic fingertip models for the primate capable of predicting mechanoreceptor response. However, as the advent of mechanoreceptor response in primates occurs at small strains [22], we concentrate on developing a viscoelastic model in these deformation ranges. Our experiments (Sec. 2.3) show that for our applied indentation depths, the response of primate skin tissue can be assumed linearly viscoelastic. To model the viscoelastic tissue, the total stress is defined as follows:

$$\sigma(t) = \sigma_0(t) + \int_0^t \dot{g}(\tau) \sigma_0(t - \tau) d\tau \quad (6)$$

where t is the time, and $g(t)$ is the stress relaxation function. The stress relaxation function is defined using a Prony series [23]

$$g(t) = 1 - \sum_{i=1}^N g_i (1 - e^{-t/\tau_i}) \quad (7)$$

where g_i and τ_i are stress relaxation parameters, and N is the number of terms used in the Prony series to define the relaxation function.

To calibrate the model, we simulated our indentation experiments with a cylindrical indenter (Sec. 2.3) in ADINA. The multi-layer model of the primate fingertip, shown in Fig. 9, was indented with a circular indenter of diameter 0.5 mm. The model was meshed using 8912 20-node solid brick elements and consisted of 315,101 nodes. The mesh was refined near the indenter as shown in Fig. 9(a) due to the small indenter size. For the boundary conditions, the indenter and bone were modeled as rigid and the nail of the finger was kept fixed as in our experiments. Similar to our experiment, the indentation consisted of a ramp (5 mm/s) up to 0.2 mm followed by a hold of 3 s after which the indenter was retracted completely. We computed the net contact force on the cylindrical indenter versus time for each simulation and calibrated two different models with our experimental data: (a) all the layers being linearly viscoelastic (we use a two term Prony series) and (b) the outer most layer being linearly elastic (with Poisson's ratio 0.48) and the rest viscoelastic (two term Prony series). The simulation parameters were optimized to fit the experimental data with results shown in Fig. 10(a). Both the homogeneous viscoelastic model and the elastic-viscoelastic model were able to fit with the experimental data well.

Fig. 9 (a) The model simulating a step indentation to a cylindrical tipped indenter similar to experiments performed in Sec. 2.3 and (b) the deformed mesh of the primate finger indented to 0.2 mm after 1 s. Effective stress is shown in megapascal.

Fig. 10 (a) The calibration curve used to estimate the viscoelastic properties of the primate fingertip model. Both the multilayer model as well as the homogeneous viscoelastic model could fit the force response data. (b) The surface deflection profiles of the calibrated multilayer model and homogeneous viscoelastic model showing that only the multilayer model is able to predict the surface deflection due to a line load as observed in experiment (Srinivasan [11]).

To validate the models, we simulated the line load experiment performed by Srinivasan [11]. A flat rigid indenter of width 0.02 mm and length 5 mm along with the fingertip was modeled in ADINA. A schematic of the model is shown in Fig. 8. The boundary conditions were the same as for the cylindrical indenter. The model was meshed using 10,820 20-node solid brick elements and consisted of 287,127 total nodes. Both the calibrated models were subjected to the line load simulation to a depth of 0.5 mm. The indentation consisted of a ramp (at 0.5 mm/s) to 0.5 mm followed by a hold of 3 s. The surface deflection along the main axis of the distal phalanx was computed after 3 s (to allow for stress relaxation). Finally, we compared the surface deflection of the two calibrated models with that of experimental results of Srinivasan [11]. The calibrated homogeneous model was able to capture the temporal response of the step indentation experiment but it was not able to capture the spatial surface deflection profile of indentation with a line load. The multilayer model, on the other hand, was found to be able to capture both the line load surface deflection as well as the temporal response to step indentation with a cylinder.

4.1 Relation of the SA Fiber Dynamic Spike Rate With Strain Energy Density Using Developed Models. There is a wealth of neurophysiological data available in literature

characterizing mechanoreceptor response to mechanical stimulation on the finger pad. Knibestöl [4,5] classified the response of mechanoreceptor afferent fibers in humans by recording its neurophysiological response to “ramp and hold” stimulus (force or displacement controlled vertical indentation on the finger pads with blunt probes). The fibers were classified as SA Merkel, rapidly adapting Meissner (RA), and Pacinian (PC) receptors based on their response to this ramp and hold stimulus. Receptors that responded only to the ramp part of the stimulus were classified as RA and PC, while the receptors that responded both to the ramp and the hold were classified as SA. The SA receptors were further subdivided into type-I and type-II based on the size of their receptor fields. These studies, however, did not take into account the shape of the indenter and thus could not say anything about the mechanoreceptors ability to discriminate shape. Phillips and Johnson [1] used stimuli consisting of gratings, bars, and edges to show that SA receptors in primates are particularly sensitive to edges. The study of the response of RA and SA receptors by Srinivasan and LaMotte [2] to sinusoidal shaped indenters as well as to cylinders of different radii showed that the surface curvature as a function of distance along the surface is a relevant parameter in discriminating shape during tactile sensing. The depth of indentation and change in curvature of skin surface, due to an applied

Fig. 11 (a) The temporal sequence of action potentials of SA afferents in response to indentation with a flat plate (data from Srinivasan and LaMotte [2]). Five independent trials are shown where each vertical spike represents an individual action potential: (b) The comparison of the strain energy density at a mechanoreceptor location with the average spike frequency (taken with 70 bins) versus time and (c) the calibrated and validated multilayered model indented with a flat plate. Effective stress is shown in megapascal.

indentation, is also shown to be represented in SA responses while rate of change of curvature of the skin surface is represented in both RAs and SAs.

2D and 3D numeric models of the primate fingertip have been used to predict SA mechanoreceptor response [6,8]. These studies showed that the strain energy density at a mechanoreceptor location matched well with the static firing rate of SA fibers [1]. However, these studies only looked at the static firing rate of the SA fiber. To show the utility of our model, we used the calibrated multilayered viscoelastic model to study the relation between the strain energy density at the mechanoreceptor location to the dynamic firing rate of the SA fiber from the literature [2]. Figure 11 shows the response of SA fibers to indentation with a flat plate. The neurophysiological response of an SA fiber to an indentation by a flat plate to a depth of 0.5 mm at a rate of 5 mm/s and held for 2.3 s is shown in Fig. 11(a). This figure, which is reproduced from Srinivasan and LaMotte [2], shows the response of four independent trials where each vertical tick represents an action potential. We simulated this indentation experiment using our calibrated multilayer model and computed the strain energy density at a mechanoreceptor location at each time step. The decay in the calculated strain energy density matched well with the previously observed decay in the average SA spike rate (Fig. 11(b)), suggesting that the dynamic spike rate may be related to the local strain energy density at the mechanoreceptor location at each time step.

5 Discussion

This study serves to add to the quantitative understanding of the mechanisms of touch by exploring the role of skin viscoelasticity in the adaptation rates of SA-I mechanoreceptors. As noted in Secs. 1 and 4.1, a wealth of neurophysiological data is available in literature where primate mechanoreceptor response has been characterized to various types of mechanical stimuli on the finger pad [1–5] along with considerable progress made in modeling these responses [6–10]. However, these studies modeled the steady-state mechanoreceptor response (or modeled the average firing rate over a fixed time window in the dynamic phase [10]) and limited literature existed that characterized the transient firing rate of mechanoreceptors during the dynamic stimulation phase. This paper contributes toward bridging this gap by studying the behavior of the mechanoreceptor in its dynamic phase along with its relation to the viscoelastic nature of the surrounding tissue. This

section discusses the rationale behind various choices made in the study, its limitations and highlights avenues for future work.

To build an accurate 3D viscoelastic finite element model of the primate finger capable of studying the dynamic behavior of mechanoreceptors, we needed experimental data to calibrate and validate the models. We decided to use already published data [11] to validate the models but collected new data to calibrate it. For our experiments, we characterized the bulk viscoelastic properties of the primate fingertip via a series of step indentation experiments on different fingers of the primate as described in Sec. 2.3. These experiments gave us 86 independent force curves taken at four different indentation depths across five fingers of the primate. Before using this data to calibrate the 3D finite element model, we wanted to understand its repeatability and variability. If these 86 independent force curves showed finger-to-finger variation, we needed to consider this when using it to calibrate the model.

Thus, our first step was to fit the experimental data via empirical viscoelastic models to study its variability and repeatability. Our approach in modeling has been to let data dictate the choice of the model by starting with the simplest model and adding complexity only when the simple model is not able to predict the data. The choice of a two dashpot, three spring Maxwell–Weichert model was driven by this principle. Analyzing the data from our indentation experiments (via the commercial data analysis software ORIGIN by Origin Lab Corporation, Northampton, MA), we found that a two term exponential time series with three coefficients fit the force curve data best. The Maxwell–Weichert element was thus constructed using two dissipative elements (dashpots) and three springs each having its own time constant. Incidentally, this result also matches with the available literature [24], where a five-parameter Maxwell–Weichert element was found to best represent the viscoelastic behavior of biological tissues.

Our fitted data (Fig. 6) show that for our given indentation ranges (200–800 μm), there is not much finger-to-finger variation in the data and in all the fitted model parameters. The data also show two distinct time constants telling us that the viscoelastic behavior in the short term (<3 s) follows a viscous liquid like behavior (fast stress relaxation with small time constant) while in the longer term (>3 s) unlike a viscous liquid with small time constant, the decay slows and the stress does not entirely reduce to zero instead slows toward an asymptotic value governed by the springs in the model. Thus, the purpose of using the

Maxwell–Weichert model was to show repeatability in our experimental data and to determine if there is any finger-to-finger variation that would need to be considered when building the finite element model.

We followed the same approach with our choice of two three-dimensional viscoelastic finite element models of the primate fingertip. We started with the simplest model (a homogeneous viscoelastic model) and decided to add a layer of complication (elastic epidermis with a viscoelastic core) only when the simplest model could not match with available data. As stated in Sec. 4, we used two sets of independent data to compare the models; force curves from our experiments in Sec. 2 for calibration and surface deflection curves from the literature [11] for validation and found the multilayer finite element model with an elastic epidermis and viscoelastic core to best fit both these sets of data. Once calibrated, this gave us the opportunity to study the dynamic relationship between mechanical stimulation of the skin and mechanoreceptor response. In turn, we found that the predicted decay in the strain energy density at the mechanoreceptor location matched well with the decay in the average SA spike rate showing that the viscoelastic tissue surrounding the mechanoreceptor has a role in regulating adaptation rates of SA-I mechanoreceptors.

There are multiple avenues for future work to expand our understanding of the dynamic phase of mechanoreceptor response. In this study, we have looked at the viscoelastic behavior of primate finger pad in response to small indentation depths (200–800 μm), where the finger pad tissue exhibits linear viscoelastic behavior. However, for larger indentation depths this assumption may not hold and thus one area of study is to characterize the same behavior for larger length scales. The current study used available neurophysiological data from a flat plate indenter [2] and all the experiments we did not vary the rate of indentation when characterizing our models. The role of indenter shape in the regulation of the dynamic firing rate of mechanoreceptors along with understanding if the rate of adaptation of SA-I mechanoreceptors in the dynamic phase is regulated the indentation rate of the indenter on the finger pad is another interesting avenue for future work.

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References

- [1] Phillips, J. R., and Johnson, K. O., 1981, "Tactile Spatial Resolution. II. Neural Representation of Bars, Edges, and Gratings in Monkey Primary Afferents," *J. Neurophysiol.*, **46**(6), pp. 1192–1203.
- [2] Srinivasan, M. A., and LaMotte, R. H., 1991, "Encoding of Shape in the Responses of Cutaneous Mechanoreceptors," *Information Processing in the Somatosensory System* (Wenner-Gren International Symposium Series), O. Franzen, and J. Westman, eds., Macmillan Press, London, UK.
- [3] Phillips, J. R., and Johnson, K. O., 1981, "Tactile Spatial Resolution. III. A Continuum Mechanics Model of Skin Predicting Mechanoreceptor Responses to Bars, Edges, and Gratings," *J. Neurophysiol.*, **46**(6), pp. 1204–1225.
- [4] Knibestöl, M., 1973, "Stimulus-Response Functions of Rapidly Adapting Mechanoreceptors in the Human Glabrous Skin Area," *J. Physiol.*, **232**(3), pp. 427–452.
- [5] Knibestöl, M., 1975, "Stimulus-Response Functions of Slowly Adapting Mechanoreceptors in the Human Glabrous Skin Area," *J. Physiol.*, **245**(1), pp. 63–80.
- [6] Srinivasan, M. A., and Dandekar, K., 1996, "An Investigation of the Mechanics of Tactile Sense Using Two-Dimensional Models of the Primate Fingertip," *ASME J. Biomech. Eng.*, **118**(1), pp. 48–55.
- [7] Maeno, T., and Kobayashi, K., 1998, "FE Analysis of the Dynamic Characteristics of the Human Finger Pad With Objects With/Without Surface Roughness," *ASME Dynamic Systems and Control Division, DSC-64*, Anaheim, CA, pp. 279–286.
- [8] Dandekar, K., Raju, B. I., and Srinivasan, M. A., 2003, "3-D Finite-Element Models of Human and Monkey Fingertips to Investigate the Mechanics of Tactile Sense," *ASME J. Biomech. Eng.*, **125**(5), pp. 682–691.
- [9] Lesniak, D. R., and Gerling, G. J., 2009, "Predicting SA-I Mechanoreceptor Spike Times With a Skin-Neuron Model," *Math. Biosci.*, **220**(1), pp. 15–23.
- [10] Gerling, G., Rivest, I., Lesniak, D., Scanlon, J., and Wan, L., 2014, "Validating a Population Model of Tactile Mechanotransduction of Slowly Adapting Type I Afferents at Levels of Skin Mechanics, Single-Unit Response and Psychophysics," *IEEE Trans. Haptics*, **7**(2), pp. 216–228.
- [11] Srinivasan, M. A., 1989, "Surface Deflection of Primate Fingertip Under Line Load," *J. Biomech.*, **22**(4), pp. 343–349.
- [12] Gerling, G. J., 2010, "SA-I Mechanoreceptor Position in Fingertip Skin May Impact Sensitivity to Edge Stimuli," *Appl. Bionics Biomech.*, **7**(1), pp. 19–29.
- [13] Wagner, M., Gerling, G. J., and Scanlon, J., 2008, "Validation of a 3-D Finite Element Human Fingerpad Model Composed of Anatomically Accurate Tissue Layers," 2008 *IEEE Haptic Interfaces for Virtual Environment and Teleoperator Systems*, Reno, NV, Mar. 13–14, pp. 101–105.
- [14] Serina, E. R., Mote, C. D., and Rempel, D., 1997, "Force Response of the Fingertip Pulp to Repeated Compression—Effects of Loading Rate, Loading Angle and Anthropometry," *J. Biomech.*, **30**(10), pp. 1035–1040.
- [15] Maeno, T., Kobayashi, K., and Yamazaki, N., 1998, "Relationship Between the Structure of Human Finger Tissue and the Location of Tactile Receptors," *Bull. JSME Int. J.*, **41**(1), pp. 94–100.
- [16] Pawluk, D. T. V., and Howe, R. D., 1999, "Dynamic Contact of the Human Fingerpad Against a Flat Surface," *ASME J. Biomech. Eng.*, **121**(6), pp. 605–611.
- [17] Wu, J. Z., Welcome, D. E., and Dong, R. G., 2006, "Three-Dimensional Finite Element Simulations of the Mechanical Response of the Fingertip to Static and Dynamic Compressions," *Comput. Methods Biomech. Biomed. Eng.*, **9**(1), pp. 55–63.
- [18] Pelli, D. G., 1997, "The VideoToolbox Software for Visual Psychophysics: Transforming Numbers Into Movies," *Spat. Vision*, **10**(4), pp. 437–442.
- [19] Meghani, S., 2004, "An Extensible Software Library for Developing Tactile Perception Experiments," M. Eng. and S.B. thesis, Department of Electrical Engineering and Computer Science, Massachusetts Institute of Technology, Cambridge, MA.
- [20] OriginLab, "Origin by Origin Lab Nonlinear Fitting Online Documentation," OriginLab Corporation, Northampton, MA, <http://www.originlab.com/doc/Origin-Help/Fitting-Nonlinear-Model>
- [21] Wan, A. W., 1994, "Biaxial Tension Test of Human Skin in vivo," *Biomed. Mater. Eng.*, **4**(7), pp. 473–486.
- [22] LaMotte, R. H., and Srinivasan, M. A., 1991, "Surface Microgeometry: Tactile Perception and Neural Encoding," *Information Processing in the Somatosensory System* (Wenner-Gren International Symposium Series), O. Franzen, and J. Westman, eds., Macmillan Press, London, UK.
- [23] Tschoegl, N. W., 1989, *The Phenomenological Theory of Linear Viscoelastic Behavior: An Introduction*, Springer, New York.
- [24] Wang, X., Schoen, J. A., and Rentschler, M. E., 2013, "A Quantitative Comparison of Soft Tissue Compressive Viscoelastic Model Accuracy," *J. Mech. Behav. Biomed. Mater.*, **20**, pp. 126–136.