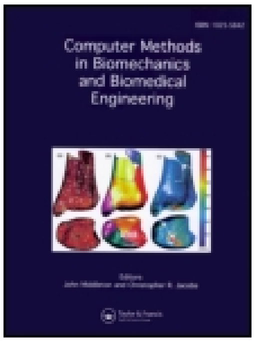

Computer Methods in Biomechanics and Biomedical Engineering

ISSN: 1025-5842 (Print) 1476-8259 (Online) Journal homepage: https://www.tandfonline.com/loi/gcmb20

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Computer Methods in Biomechanics and Biomedical Engineering,
Vol. 9, No. 1, February 2006, 55–63

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Three-dimensional finite element simulations of the mechanical response of the fingertip to static and dynamic compressions

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(Received 1 November 2005; in final form 20 January 2006)

The analysis of the mechanics of the contact interactions of fingers/handle and the stress/strain distributions in the soft tissues in the fingertip is essential to optimize design of tools to reduce many occupation-related hand disorders. In the present study, a three-dimensional (3D) finite element (FE) model for the fingertip is proposed to simulate the nonlinear and time-dependent responses of a fingertip to static and dynamic loadings. The proposed FE model incorporates the essential anatomical structures of a finger: skin layers (outer and inner skins), subcutaneous tissue, bone and nail. The soft tissues (inner skin and subcutaneous tissue) are considered to be nonlinearly viscoelastic, while the hard tissues (outer skin, bone and nail) are considered to be linearly elastic. The proposed model has been used to simulate two loading scenarios: (a) the contact interactions between the fingertip and a flat surface and (b) the indentation of the fingerpad via a sharp wedge. For case (a), the predicted force/displacement relationships and time-dependent force responses are compared with the published experimental data; for case (b), the skin surface deflection profiles were predicted and compared with the published experimental observations. Furthermore, for both cases, the time-dependent stress/strain distributions within the tissues of the fingertip were calculated. The good agreement between the model predictions and the experimental observations indicates that the present model is capable of predicting realistic time-dependent force/displacement responses and stress/strain distributions in the soft tissues for dynamic loading conditions.

Keywords: Finite element model; Soft tissue mechanics; Fingertip; Contact pressure; Viscoelasticity

1. Introduction

For any tools, handles serve as an interface between operators and machines. The contact interactions between hands and handles may interfere with manipulations (Birznieks *et al.* 1998) and loading distributions in musculoskeletal systems (Gurram *et al.* 1995) during operations. Many occupation-related disorders in hand and fingers, such as carpal tunnel syndrome and hand-arm vibration syndrome, are believed to be associated with the contact pressure between the fingers and the tool-handle (Radwin *et al.* 1987, Reidel 1995). Professional musicians, such as string instrument and piano players, may also develop hand or finger disorders (Lockwood 1989, Robinson and Kincaid 2004). Although, the mechanisms of these hand disorders are not known explicitly, experimental evidence indicates that they are mechanically mediated. High contact pressure concentration at the fingertip, which may occur while operating hand-held tools or playing instruments, can introduce excessive deformation of soft tissues, and thus initiating adaptation or degenerations of the vascular and neural sensory systems in the fingertips. Since the experimental determinations of the distributions of the stress/strain in the soft tissues are technically very difficult, theoretical or numerical analyses are used to provide essential information for such studies.

In principle, two types of models are proposed in the literature to mimic the anatomical structures of fingertips: "waterbed" and "continuum" models. In the "waterbed" models (Srinivasan 1989, Serina *et al.* 1998), a fingertip is represented by an incompressible fluid representing the subcutaneous tissue enclosed by an elastic or hyperelastic membrane representing the skin. The "waterbed" models have been applied to predict the deflection of the fingertip

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Computer Methods in Biomechanics and Biomedical Engineering
ISSN 1025-5842 print/ISSN 1476-8259 online
http://www.tandf.co.uk/journals
DOI: 10.1080/10255840600603641

surface under a static line load (Srinivasan 1989) and static force-deflection characteristics (Serina *et al.* 1998). These models, however, cannot adequately predict the stress and strain in the soft tissues. In the "continuum" fingertip models, such as those proposed in the literature (Phillips and Johnson 1981, Srinivasan and Dandekar 1996, Dandekar *et al.* 2003), the skin and subcutaneous tissues were represented by homogeneous or multi-layered elastic media. The continuum fingertip models can predict the stress/strain distributions within the tissues. In these finite element (FE) fingertip models, the skin and subcutaneous tissues were considered to be linearly elastic and time-independent, such that the nonlinear and time-dependent behaviors of the fingertips for physiological loading conditions cannot be investigated. In our previous two-dimensional (2D) FE models (Wu *et al.* 2003), the soft tissues in the fingertip were simulated using nonlinearly viscoelastic material models. However, the application of the 2D models is limited, since all realistic biomechanical problems are three-dimensional (3D) in nature.

One of the technical challenges in the FE modelling of fingertip is to obtain reliable experimental data of the soft tissues. The subcutaneous and skin tissues are known to have nonlinear and time-dependent mechanical behaviors (Wan Abas 1994, Zheng and Mak 1996, Rubin *et al.* 1998). Most published mechanical properties of the soft tissues were obtained via *in vitro* tests. The mechanical characteristics of the soft tissues are expected to be varied during *in vitro* tests, due to many factors, such as the loss of natural tension, the absence of blood flow and the disconnecting of fibrous structures in the specimen preparation. Hendriks *et al.* (2003, 2004) tested the nonlinearly elastic behaviors of human skin *in vivo* using optical coherence tomography. The nonlinear stress/strain relationships obtained in the *in vivo* tests are believed to be more relevant to the physiological conditions.

The contact interactions between fingers and handle may interfere with grasp stability, thereby affecting the manipulations of hand-held tools. For example, the grip forces applied in operation of hand-held tools (e.g. screw drivers) were found to be dependent on the length and diameter of the handles (Freund *et al.* 2002); the handle diameter of chopsticks was found to influence significantly the food-serving performance (Wu 1995, Chan 1999). All of these previous studies were performed experimentally; the contact interactions between the fingertip and contact surfaces have seldom been analyzed. From a biomechanical point-of-view, the contact interactions between the fingers and tool handle will affect operator comfort and manipulability of the hand tools. In the present study, a 3D FE model for fingertip is proposed to simulate the nonlinear and time-dependent responses of a fingertip to static and dynamic loadings. In order to simulate the realistic stress/strain distributions in the soft tissues, the essential anatomical structures of a finger are incorporated into the proposed FE model: skin layers (dermis and epidermis), subcutaneous tissue, bone and nail. The contact interactions between a fingertip and two different objects for the continuous and transient dynamic loading are investigated using the proposed model, and the simulation results are compared with the published experimental data.

2. Methods

2.1 FE model

The fingertip considered in the model is the distal phalanx, the portion from the finger distal to the distal interphalangeal (DIP) joint articulation, as shown in figure 1. The external shape of the fingertip was determined using a smooth mathematical surface fitting to the digitized fingertip shapes in a study "evaluation of deformation and contact pressure of fingertips under static compression: a pilot study" (unpublished research protocol). The fingertip surface was then scaled to the dimensions obtained in the experiments (Dequeker and Vadakethala 1979, Bruhn *et al.* 1991, Murai *et al.* 1997). The fingertip model has a length of 25 mm, a width of 20 mm and a height of 18 mm, which are considered to be representative for a typical male index finger. The fingertip was assumed to be symmetric, such that only a half of the fingertip was considered in the modelling.

The cross-sectional anatomical structures of the model were developed based on the published magnetic resonance imaging of human fingers (Bruhn *et al.* 1991). The bone was elliptical and the tissue thickness was considered to be asymmetric about the bone. The bone size and soft tissue

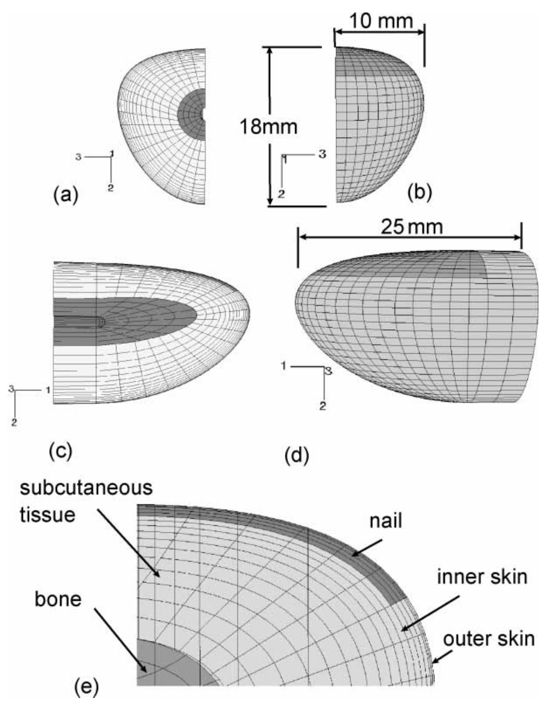

Figure 1. 3D FE fingertip model. (a) and (c) are the cross and longitudinal section, respectively. (b) and (d) are the external view from the top and side, respectively. (e) shows the detailed substructures of the model. The fingertip is approximated as symmetric, such that only a half of the fingertip is modelled. The fingertip model contains bone, nail, subcutaneous tissue, inner skin and outer skin. The fingertip section model has a width of 20 mm, a height of 18 mm and a length of 25 mm.

thickness of the model were taken from the published experimental data (Pejovic-Milic *et al.* 2002). The nail was considered to have a thickness of 0.60 mm (Baran 2004). The skin layer is composed of epidermis (44–116 μm thick) and dermis (0.8–1.2 mm thick) (Hendriks *et al.* 2003, 2004, Agache 2004). The epidermis is further divided into stratum corneum (SC) (14–36 μm thick) and viable epidermis (30–80 μm thick). The SC layer is linearly elastic and much stiffer than the dermis and has a Young's modulus around 12 MPa (Takenouchi *et al.* 1986). We did not find any published data for the mechanical properties of the viable epidermis. Since, the viable epidermis is a transition from SC to dermis in anatomical structure, it is reasonable to assume that the stiffness of the viable epidermis is between those of SC and dermis. In our FE model, the skin is assumed to be composed of two layers: the outer skin (100 μm thick) and inner skin (1.2 mm thick). The outer skin layer, composed of SC and part of epidermis, is considered to be linearly elastic. The Young's modulus and Poisson's ratio of the outer skin were assumed to be 2.0 MPa and 0.3 (Agache 2004), respectively. The inner skin, composed of dermis and part of epidermis, is considered to be linearly viscous and nonlinearly elastic. The *in vivo* test data of human skin (Hendriks *et al.* 2003, 2004) were applied to determine the nonlinearly elastic characteristics of the inner skin.

The material properties of the bone and nail were assumed to be linearly elastic. Based on the published experimental data (Yamada 1970), the Young's moduli of the bone and nail were assumed to be 17.0 and 170.0 MPa, respectively, while Poisson's ratio was assumed to be 0.30 for both. The contact object (i.e. the flat surface and wedge) was considered to be composed of PVC polymer with a Young's modulus of 2.8 GPa and Poisson's ratio of 0.30.

2.2 Constitutive equations for the inner skin and subcutaneous tissue

The constitutive equations of the inner skin and subcutaneous tissue are similar to those used in our previous study (Wu *et al.* 2003). The total tissue stress ($\tilde{\sigma}$) is assumed to be composed of elastic ($\tilde{\sigma}^0$) and viscous ($\tilde{\sigma}^v$) stress components, such that:

$$\begin{aligned}\tilde{\sigma}(t) &= \tilde{\sigma}^0(t) + \tilde{\sigma}^v(t) = \tilde{\sigma}^0(t) + \int_0^t \frac{\dot{G}(\tau)}{G_0}\tilde{\sigma}^0(t-\tau)\,\mathrm{d}\tau \\ &= \tilde{\sigma}^0(t) + \int_0^t \dot{g}(\tau)\tilde{\sigma}^0(t-\tau)\,\mathrm{d}\tau \qquad (1)\end{aligned}$$

where t is time. The stress relaxation function is defined using the Prony series (Tschoegl 1989):

$$g(t) = \frac{G(t)}{G_0} = \left[1 - \sum_{i=1}^{N_G} g_i\left(1 - \mathrm{e}^{-\frac{t}{\tau_i}}\right)\right] \qquad (2)$$

where G_0 and $G(t)$ are the instantaneous and time-dependent moduli, respectively; g_i and τ_i ($i = 1, 2, \ldots N_G$) are stress relaxation parameters; N_G is the number of terms used in the stress relaxation function.

The elastic deformation behaviors of both inner skin and subcutaneous tissues are simulated using the polynomial model which is defined using a function of strain energy density per unit volume:

$$U = \sum_{i+j=1}^{N} C_{ij}(\bar{I}_1 - 3)^i(\bar{I}_2 - 3)^j + \sum_{i=1}^{N}\frac{1}{D_i}(J-1)^{2i} \qquad (3)$$

where $\bar{I}_1$ and $\bar{I}_2$ are the first and second deviatoric strain invariants, respectively; J is the elastic volume ratio; N, D_i and C_{ij} are the material parameters.

The subcutaneous tissue is assumed to have similar nonlinear characteristics to the inner skin. The material parameters of the inner skin are determined using the *in vivo* tests (Hendriks *et al.* 2003, 2004), while those of the subcutaneous tissue will be determined using a test and try procedure by fitting the predicted mechanical responses of the fingertip to the experimental data (Serina *et al.* 1997, Wu *et al.* 2003). In order to minimize the number of the parameters to be determined in the numerical tests, we assume that the stiffness ratio between the subcutaneous tissue and inner skin is constant, i.e. $\sigma_{\text{sub}} = r\sigma_{\text{skin}}$ for given strain, where r is the stiffness ratio, σ_{sub} and σ_{skin} are the stress in subcutaneous tissue and inner skin in uniaxial tests, respectively. The stiffness ratio r is the only parameter that is to be determined by calibrating the model predictions to the published experimental data. The elastic stress/strain relationships of the inner skin and subcutaneous tissues used in the simulations are demonstrated in figure 2. The figure shows the stress/strain relationships of the subcutaneous tissues using stiffness parameters $r = 1/16$, 1/8, 1/4 and 1/2. A two-term ($N = 2$) equation [equation (3)] was used to fit the experimental data. The material parameters of

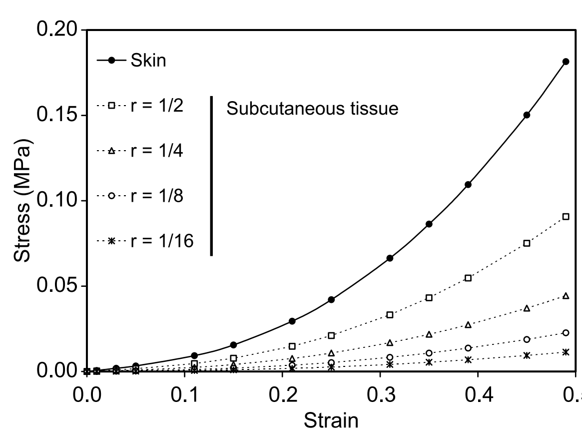

Figure 2. Nonlinear elastic properties of the inner skin and subcutaneous tissues. r is the stiffness ratio of the subcutaneous tissue to the inner skin. The stress/strain relationship for the inner skin is assumed to be fixed, while that for the subcutaneous tissue has been varied ($r = 1/16$, 1/8, 1/4 and 1/2) in the calculations to achieve the best fitting of the predicted force or deformation to the experimental data.

Table 1. Material parameters of the inner skin and subcutaneous tissues.

(a) Viscoelastic parameters for the soft tissues	i	1	2
Inner skin	g_i	8.64E − 2	2.14E − 1
	τ_i (s)	2.12E − 1	8.85
Subcutaneous tissue	g_i	2.57E − 1	3.83E − 1
	τ_i (s)	2.23E − 1	4.68

(b) Elastic parameters for the soft tissues	C_{10} (MPa)	C_{01} (MPa)	C_{11} (MPa)	C_{20} (MPa)	C_{02} (MPa)	D_1 (MPa^{-1})
Inner skin	2.34E − 03	5.42E − 03	−0.262	0.239	7.47E − 2	13.3
Subcutaneous tissue						
$r = 1/16$	1.50E − 4	3.36E − 4	−1.64E − 2	1.49E − 2	4.66E − 3	213.0
$r = 1/8$	3.00E − 4	6.71E − 4	−3.27E − 2	2.98E − 2	9.33E − 3	106.5
$r = 1/4$	5.97E − 4	1.34E − 3	−6.55E − 2	5.96E − 2	1.87E − 2	53.3
$r = 1/2$	1.19E − 3	2.70E − 3	−0.131	0.119	3.73E − 2	26.6

The parameters of the inner skin were obtained by fitting the constitutive model [equation (3)] to the *in vivo* test data of human skin (Hendriks *et al.* 2003, 2004). Both inner skin and subcutaneous tissue were considered to be nearly incompressible.

the soft tissues used in the simulations are listed in table 1.

2.3 Numerical simulations

Two loading scenarios will be simulated in the present study (figure 3): (a) the contact interaction between the fingerpad and a flat surface and (b) the indentation of the fingerpad via a sharp wedge. These two loading cases are representative for many practical problems, such as the finger/handle contacts for hand tools, and finger/string interactions for musical (string) instruments. The numerical tests were performed using displacement-controlled protocols, in which the fingertip was pressed against the contact object via prescribed time-histories. In scenario (a), we have studied the nonlinear elastic characteristics of the system by using a very slow loading rate (0.1 mm/s), and the time-dependent responses of the system by using different loading/unloading time-histories. In scenario (b), we have studied surface deformation profiles and time-dependent force responses of the fingertip by using loading rates from 0.1 to 2.0 mm/s. In both cases, the friction between the skin surface and the contacting object was assumed to be negligible.

The simulations were performed using the commercial FE software package Abaqus (version 6.5). The fingertip model contains 4,414 20-node quadratic elements and 15,872 nodes.

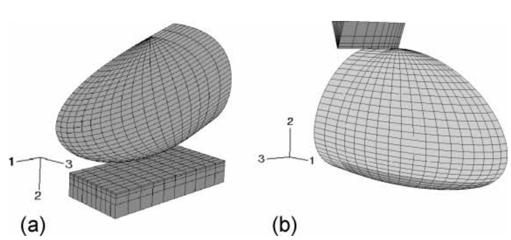

Figure 3. FE models of the contact of the fingertip with two different objects. (a) The contact interaction between the fingerpad and a flat surface. (b) The indentation of the fingerpad via a line load.

3. Results

For the contact of the fingertip with a flat surface [Scenario (a)], the distributions of the displacement, stress and strain in the tissues of the fingertip have been predicted. Typical distributions of the vertical displacement in the tissues of the fingertip are depicted in figure 4. The simulation results shown were obtained by using the elastic stiffness ratio $r = 1/4$ for the subcutaneous tissues. The model predictions for the static force/displacement relationships for scenario (a) are compared with the experimental data reported by Serina *et al.* (1997) and Wu *et al.* (2003) (figure 5). These numerical tests were performed using such a slow loading rate (0.1 mm/s) that the effects of the material viscosity become negligible. The force/displacement relationships of the fingertip were predicted by varying the stiffness of the subcutaneous tissues at four levels (i.e. $r = 1/16$, 1/8, 1/4 and 1/2). It is seen that the contact stiffness of the fingertip and the plate increases with increasing stiffness ratio r. The predicted force/displacement relationships using $r = 1/8$ and $r = 1/4$ fit best to the experimental data by Serina *et al.* (1997) and Wu *et al.* (2003), respectively. Considering the variations of the test data, the fitting of the model predictions to the experimental data is satisfactory. The numerical tests demonstrate that the individual test results can be well simulated by varying a single parameter r.

The dynamic contact interactions between the fingertip and the flat surface were further investigated via a loading/unloading protocol, as shown in figure 6. In this series of tests, the fingertip was first compressed to 2 mm at four different speeds (i.e. 5.0, 0.4, 0.2 and 0.1 mm/s as labelled A, B, C and D, respectively, in figure 6), held constant for 30 s, unloaded to a reduced compression of 1 mm at a speed of 1 mm/s, and the compression displacement (1 mm) was then held constant for another 30 s. The predicted time-histories of the force responses are compared with the experimental measurements (Wu *et al.* 2003) for these four tests, as shown in

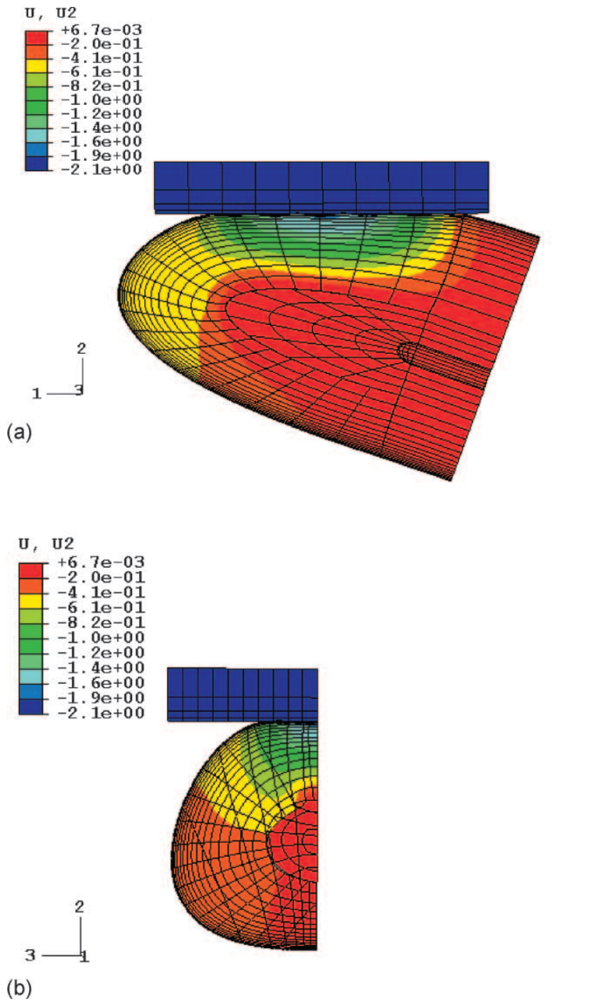

Figure 4. The predicted distributions of the displacement in the vertical direction ($U2$ in mm) in the soft tissues of the fingertip pressed against a flat plate. (a) and (b) shows the longitudinal and cross sections of the fingertip, respectively. A displacement of 2.1 mm was applied on the contact plate.

figure 7. For all these four cases, the relative differences between the predicted peak-force values and the corresponding averaged experimental data are less than 10%. A stiffness ratio of $r = 1/8$ was applied in all four numerical tests.

For the indentation of the fingerpad via a line load [Scenario (b)], the distributions of the displacement, stress and strain in the tissues of the fingertip have been predicted. Typical results of the distribution of the vertical displacement in the tissues of the fingertip are depicted in figure 8. The surface deflections as a function of the distance from the load were predicted and compared with the experimental data reported by Srinivasan (1989) (figure 9). These numerical tests were performed using such a slow loading rate (0.1 mm/s) that the effects of the material viscosity become negligible. The surface deformation profiles were calculated by using the stiffness of the subcutaneous tissues at four levels (i.e. $r = 1/16$, 1/8, 1/4 and 1/2). The optimized fitting of the predicted deflection profiles to the experimental measurements

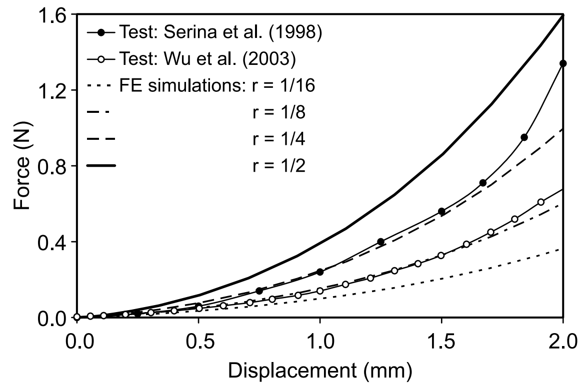

Figure 5. The predicted static force–displacement relationships for the contact of the fingertip with a flat surface. The calculations were performed using four different tissue stiffness ratios ($r = 1/16$, 1/8, 1/4 and 1/2). The force/displacement curves predicted using $r = 1/8$ and 1/4 fit to the experimental data by Serina *et al.* (1998) and Wu *et al.* (2003), respectively. The fingertip was pressed at very slow rate (0.1 mm/s) in these tests.

(with errors less than 10%) is achieved by using relative stiffness parameters between 1/4 and 1/2. The distributions of the vertical displacement in the cross-section of the fingertip for a typical case ($r = 1/4$) are shown in figure 8. It is seen that the tissue deformation gradients are concentrated at the location of the skin/wedge contact.

In order to investigate the time-dependent contact force responses of the fingertip, the loading edge was placed onto the fingertip at five different rates (i.e. 0.1, 0.2, 0.5, 1.0 2.0 mm/s), the corresponding force/displacement relationships of the fingertip were predicted, as shown in figure 10. The simulations were performed using the stiffness ratio $r = 1/4$ for the subcutaneous tissue. The simulation results demonstrated that the contact stiffness of the fingertip increases slightly with increasing loading rate.

4. Discussion and conclusion

The contact interactions between fingers and hand-held tools interfere with the tool manipulability and comfort of the operators. Since the contact pressure has been

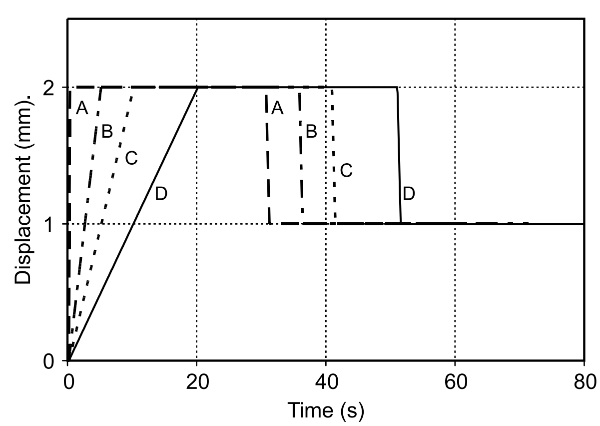

Figure 6. The displacement time-histories of the contact plate used in the study. The flat contact plate was prescribed to four (A, B, C and D) different displacement time-histories in the simulations.

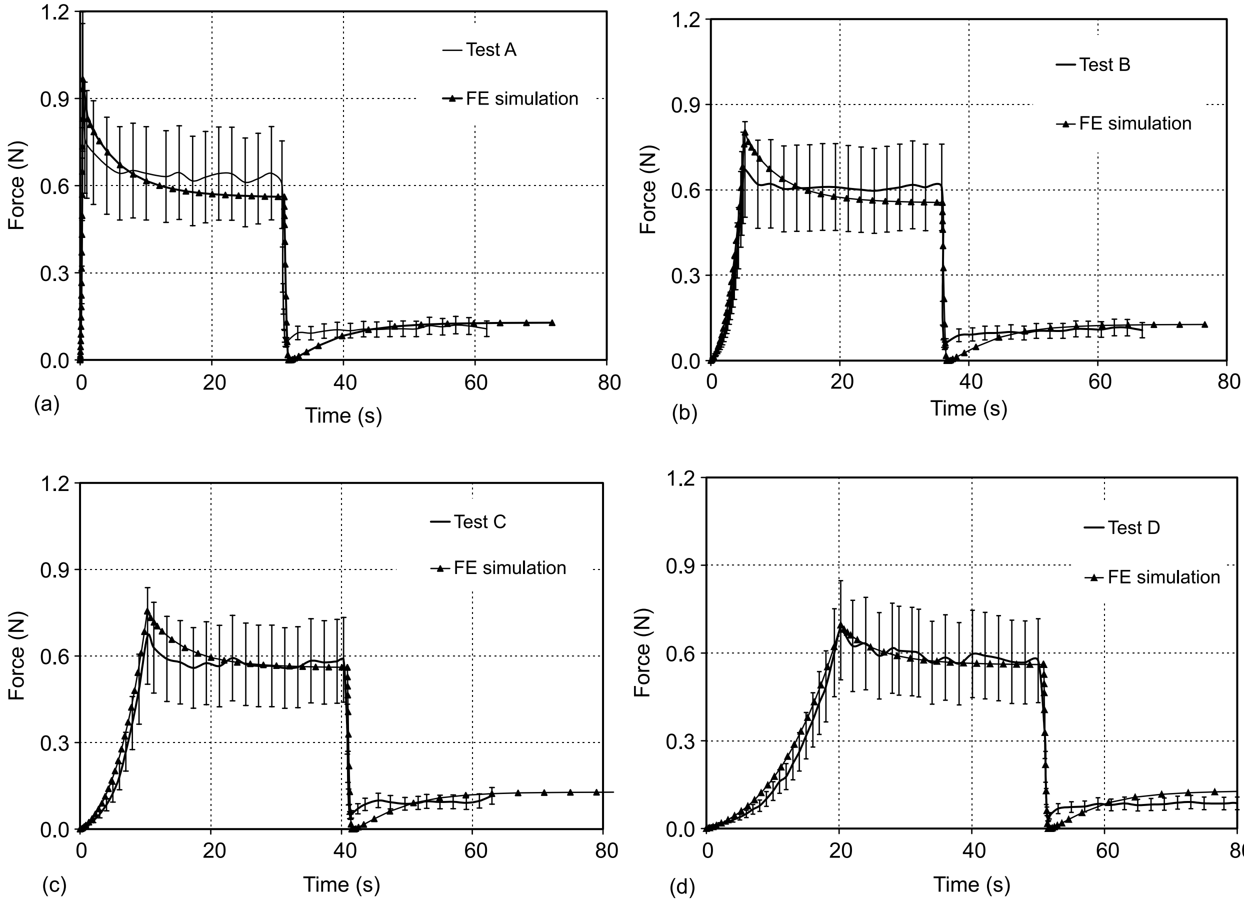

Figure 7. The comparison of the predicted time-dependent force responses with the experimental data (Wu *et al.* 2003). The figures show that the predicted force responses fit well with the experimental data (Wu *et al.* 2003).

associated with many occupational hand-finger disorders, it has been used as factor of risk assessment. Engineers and ergonomic designers have put forth considerable effort in designing tool handle to reduce the fatigue and to improve comfort of the operators. All these previous studies have been conducted experimentally. From a biomechanical point-of-view, the comfort and manipulability of a hand tool is associated with the finger/tool contact interactions, i.e. contact pressure, contact force, local tissue deformations, etc. The contact interactions influence the stress/strain distributions in the soft tissues, which could in turn trigger the adaptation of the vascular and neural systems in the fingertip. Numerous factors, such as the contact angle, touching speed, the shape and curvature of the contact surface and the materials of the contact body, will influence the contact interactions. The knowledge of the effects of these contact conditions on the contact interaction of fingers/tool can only be obtained via theoretical analyzes to date. A model based upon a simplified anatomical representation of the fingertip incorporating time-dependent, nonlinear characteristics of the soft tissues could thus provide considerable insight into the mechanism of the initiation and development of many occupational hand disorders and help optimize the tool design to reduce occupational musculoskeletal disorders in operators' hand/fingers.

In the previous study (Dandekar *et al.* 2003), the material properties of the soft tissues were assumed as linearly elastic and the elastic moduli of the tissues were obtained by fitting the predicted skin deformation to the test data. Although, these models were able to achieve an optimized fitting to the surface deformation observed in the experiments, they would not be able to predict reliable stress/strain distributions inside the soft tissues and force/displacement relationship of the fingertip, because of the oversimplified material models of the soft tissues. Compared with the previous FE model (Dandekar *et al.* 2003), the present model includes the multi-layered soft tissue structures, which mimic realistic anatomies of a fingertip, and nonlinearly elastic and linearly viscoelastic properties of the soft tissues, which are based on *in vivo* tests. Therefore, the present model is believed to be capable of predicting realistic time-dependent stress/strain distributions in the soft tissues and force/displacement responses for dynamic loading conditions.

The simulated distributions of the tissue displacement for both loading scenarios (figures 4 and 8) show that the deformation gradients or strains in the soft tissues are highly concentrated at the location of the contact interface. The strain concentration patterns in cross-sections are similar [figures 4(b) vs 8(b)] for these two very different loading cases.

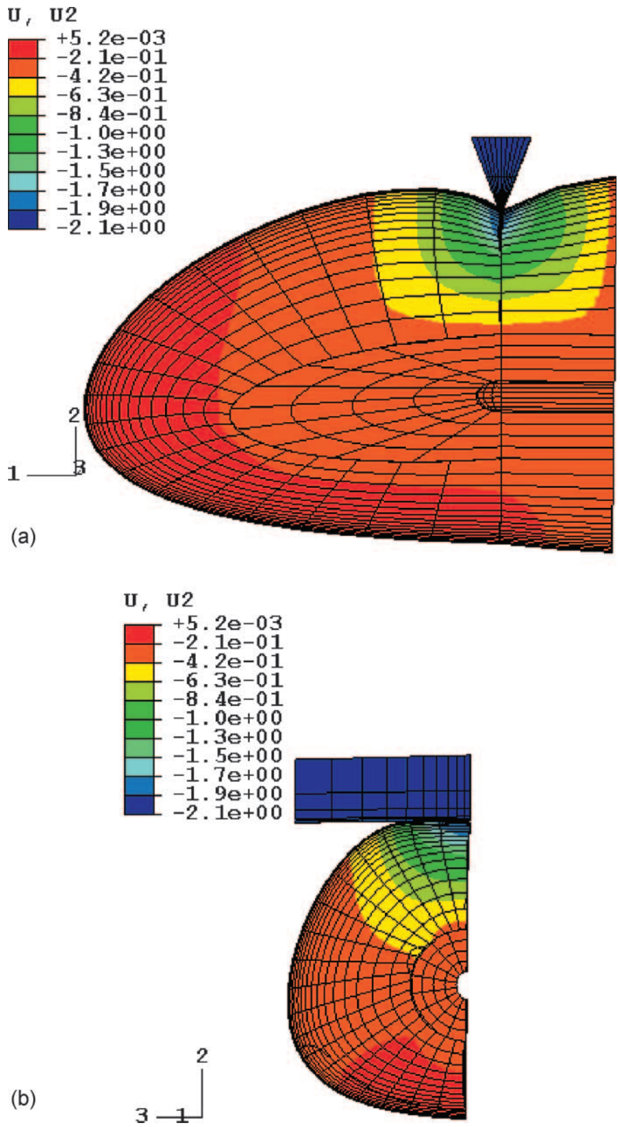

Figure 8. The predicted distributions of the displacement in the vertical direction ($U2$ in mm) in the soft tissues of the fingertip indented via a line load. (a) and (b) shows the longitudinal and cross sections of the fingertip, respectively. The fingerpad is indented by a displacement of 2.1 mm.

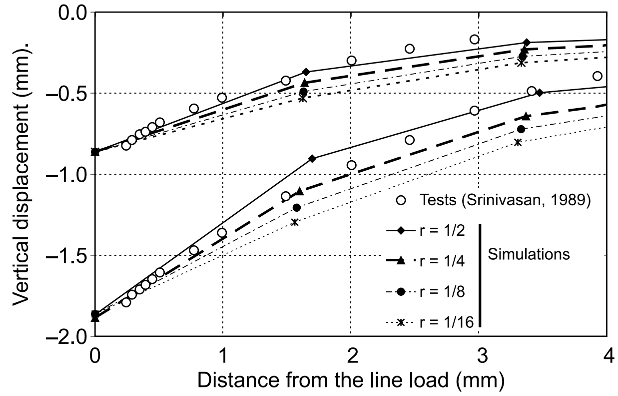

Figure 9. The predicted surface deflection profiles of the fingerpad indented via a line load. The horizontal axis is the distance from the spot of the line-load and towards the distal of the fingertip. The simulation results are compared with the experimental data by Srinivasan (1989). The figures show that the predicted deflection profiles by using $r = 1/4$ get the best fit to the experimental data. The tests were performed by applying a very slow loading rate (0.1 mm/s).

In the present study, only the distal section of an index finger was modelled, so that the technical difficulties in the modelling of the joint can be avoided. The cross section at the cut-off location was assumed to be unchanged for all simulation results presented here. The effects of the boundary conditions at the cut-off section were tested by varying from unconstrained to constrained, and were found to have negligible effects on the simulation results.

The predicted time-histories of the force responses agree well in trends with the corresponding data for the dynamic contact of the fingertip with the flat surface (figure 7). At the moment, when the compressed displacement was suddenly released from 2 to 1 mm, the model predicts that the contact force reaches about zero and then recovers to higher levels. These simulation results are a little different from the experimental observations, which showed higher minimum force values. The difference between the theoretical prediction and experiment data at that particular point is likely due to artifacts in the experiments. The small contact forces (< 5 g) observed in the experiments may be caused by the inertia of the indentation plate, while the simulations were performed using a quasi-static scheme in which the inertial effects of the contact platen were excluded.

In the present study, we have applied the best knowledge on the anatomical structures and material properties of the soft tissues to the modelling. The only material parameter that is altered to obtain the best fitting of the model prediction to the test data is the elastic properties of the subcutaneous tissues. Our simulation results indicated that the stiffness ratio of the subcutaneous tissue over the inner skin should be between 1/8 and 1/4 for the model predictions to fit the experimental data. These results are consistent with the results of soft tissue testing (Hendriks *et al.* 2003, Wu *et al.* 2005).

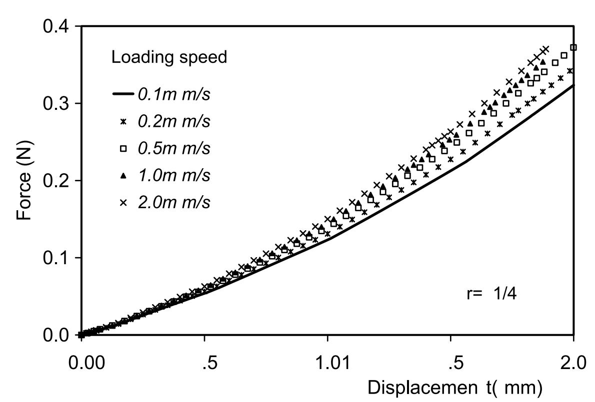

Figure 10. The predicted time-dependent force–displacement relationship of the fingertip indented via a line load. The fingerpad was indented via a line load at five different speeds from 0.1 to 2.0 mm/s. The simulation results show that the force responses of the fingertip depend on the loading speed: the stiffness of the fingertip increases with the increasing loading speed.

The present study was conducted using a quasi-static approach, i.e. the material properties of the soft tissues were considered as time-dependent, while the inertia effects of the materials during the movement were neglected. Despite the simplifying assumption, the proposed model can be applied to explore the solutions for many practical problems. Our previous studies indicated that during the impact between a fingertip and a keypad, the impact force can be decomposed into a portion associated with the inertia of the tissue mass and a portion associated with tissue deformations (Wu *et al.* 2003). The quasi-static model, as presented in this study, is applicable of predicting the deformation-dependent impact force component during the keypad strike (Wu *et al.* 2003). Since, the effects of the mass inertia on the tissue deformation is negligible, the proposed quasi-static model can also be applied to analyze the time-dependent deformations and stress/strain distributions of the soft tissues during the dynamic loadings, for example, to analyze the mechanical responses of the soft tissues in fingertip in vibrotactile test (Wu *et al.* 2003), in playing string instruments, piano and during a keypad strike (Wu *et al.* 2003), etc. Furthermore, the proposed model can be used to calculate the relationship between the statically applied force and the contact pressure in the interface of finger/power-tool-handle. The applied static force was found to influence the mechanical impedance and energy absorbtion in fingers during vibrations (Dong *et al.* 2004, 2005). The stress/strain and strain energy absorbtion in the soft tissues in vibration have been proposed to interfere with the development of the hand-arm vibration syndromes. The current quasi-static model provides an essential step towards the development of a fully dynamic finger model.

In summary, in the present study we proposed a novel 3D FE fingertip model, which contains realistic anatomic micro-structures and nonlinear viscoelastic material properties of the soft tissues. The model predictions on the fingerpad deformations and time-dependent force responses of the fingertip agree well with the published experimental observations.

Acknowledgements

The findings and conclusions in this report are those of the authors and do not necessarily represent the views of the National Institute for Occupational Safety and Health.

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