Extraction of Quasi-Linear Viscoelastic Parameters for Lower Limb Soft Tissues From Manual Indentation Experiment

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A manual indentation protocol was established to assess the quasi-linear viscoelastic (QLV) properties of lower limb soft tissues. The QLV parameters were extracted using a curve-fitting procedure on the experimental indentation data. The load-indentation responses were obtained using an ultrasound indentation apparatus with a hand-held pen-sized probe. Limb soft tissues at four sites of eight normal young subjects were tested in three body postures. Four QLV model parameters were extracted from the experimental data. The initial modulus E_0 ranged from 0.22 kPa to 58.4 kPa. The nonlinear factor E_1 ranged from 21.7 kPa to 547 kPa. The time constant τ ranged from 0.05 s to 8.93 s. The time-dependent material parameter α ranged from 0.029 to 0.277. Large variations of the parameters were noted among subjects, sites, and postures.

Introduction

The human musculoskeletal system is entirely covered by layers of soft tissues. The material properties of these soft tissues are important to how the interface pressures are developed as the epidermal surface interacts with an external body support surface such as in the designs of seating, mattresses, as well as prosthetic sockets. In addition, biomechanical assessment of skin and the underlying soft tissues on a bony substratum may also be relevant to the monitoring of some pathological conditions, such as edema, and to dermatology in general.

Residual limb evaluation is an important step in the process of prosthesis design. In addition to dimensional measurements, the biomechanical properties of the residual limb tissues are important factors in deciding the amount of cast rectification to be made in the process of prosthetic socket design in order for the prosthesis to fit the specific residual limb of an amputee. For the lack of an easy-to-use quantitative tool to assess the biomechanical properties of limb tissues in the clinic, palpation, namely feeling the stiffness of the soft tissue with fingers pushing on the skin, is widely used by prosthetists to assess the residual limb tissues. Such subjective assessments require substantial experience acquired through trial and error. The qualitative nature makes knowledge accumulation difficult and makes teaching/learning imprecise.

During the last decade, computer-aided design and computer-aided manufacturing (CAD/CAM) have been introduced to prosthetic socket design (Saunders et al., 1985; Dean and Saunders, 1985; Houston et al., 1992; Torres-Moreno et al., 1992a, Boone et al., 1992, 1994; Engsberg et al., 1992; Ellepola et al., 1993, Lilja and Öberg, 1995). CAD/CAM systems make the fabrication of the prosthetic socket much more convenient than the conventional ways. Finite element (FE) analysis has been used to predict the interface stress between the surface of the residual limb tissue and the socket as well as the internal stresses and strains in the soft tissues (Steege et al., 1987, 1988, 1992; Krouskop et al., 1987; Silver-Thorn and Childress, 1991, 1992, 1996; Brennan and Childress, 1991; Reynolds and Lord, 1988, 1992; Sanders and Daly, 1991, 1993; Zachariah and Sanders, 1996; Quesada and Skinner, 1991, 1992; Torres-Moreno et al., 1991, 1992b; Lee et al., 1993, 1995; Mak et al., 1992; Mak and Huang, 1994; Huang and Mak, 1994a; Zhang et al., 1995; Zhang and Mak, 1996). It is anticipated that FE analysis can offer the opportunity to evaluate a prosthetic design before the design is fabricated. Biomechanical properties of the residual limb tissues are important inputs to the design of prosthetic sockets. However, for the lack of an easy-to-use approach and for the paucity of nonlinear and time-dependent properties of these tissues in the literature, the complex behavior of the soft tissue properties were not well represented in current stump–socket modeling. It has been widely suggested that more accurate description of the soft tissue properties should be taken into consideration (Steege et al., 1987; Torres-Moreno, 1991; Sanders and Daly, 1993; Zhang et al., 1995; Silver-Thorn and Childress, 1996).

An indentation apparatus was developed by Schade et al. (1912) to study the changes of the creep properties of skin and subcutaneous tissues in edematous limbs. Investigations have also been conducted on the dependence of biomechanical properties of soft tissues on site, muscular contraction state, individual, age and gender, etc. (Kirk and Kvorning, 1949; Kirk and Chieffi, 1962; Lewis et al., 1965; Ziegert and Lewis, 1978; Dikstein and Hartzshtark, 1981; Bader and Bowker, 1983; Horikawa et al., 1993). Testing sites on lower limbs and forearms were usually selected in those investigations. Additional apparatus have recently been developed for similar purposes (Krouskop et al., 1987; Steege et al., 1987, 1988; Silver-Thorn and Childress, 1991, 1992; Vannah and Childress, 1988, 1996; Reynolds and Lord, 1988, 1992; Torres-Moreno et al., 1991, 1992b; Mak et al., 1994a; Ferguson-Pell, 1994; Zheng and Mak, 1995, 1996). In spite of these investigations, detailed quantitative descriptions of the biomechanical properties of limb soft tissues at different sites and with different body postures are still lacking.

In earlier investigations, indentation tests were always performed in a loading-creep-unloading sequence, and the results were analyzed in a qualitative or semi-quantitative way with some directly derived empirical parameters, such as the instantaneous indentation expressed in percentage of total indentation. Later, linear indentation solutions (Timoshenko and Goodier, 1970; Hayes et al., 1972) were used to extract the elasticity of

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soft tissues from indentation results (Steege et al., 1987; Steege and Childress, 1988, 1992; Torres-Moreno, 1991; Torres-Moreno et al., 1992b; Vannah and Childress, 1988; Reynolds, 1988; Reynolds and Lord, 1992; Mak et al., 1994a; Zheng and Mak, 1996, 1998a, b). Since the nonlinearity and nonhomogeneity of soft tissues were neglected in the extraction of these Young's moduli, the parameters obtained were often referred to as the effective Young's modulus or effective elastic modulus. Finite element analysis was also adopted by some investigators to determine the material properties of the residual limb tissues of lower limb amputees (Steege et al., 1987; Reynolds, 1988; Silver-Thorn, 1991; Vannah and Childress, 1996). Torres-Moreno (1991) measured the moduli at several different levels of indentation to demonstrate the nonlinear dependence of the soft tissue properties. Most recently, Vannah and Childress (1996) reported their approach to extract nonlinear material parameters of soft tissues using a strain energy function. However, in general, there have been very few reports on the nonlinear and time-dependent properties of limb soft tissues.

The fact that biological soft tissues exhibited time-dependent properties are well known. To address the creep-indentation behavior of articular cartilage, Coletti et al. (1972) modeled the phenomenon using a discrete spring-dashpot Kelvin–Voigt type viscoelastic law. Similar lumped-parameter modeling procedures were also employed by Hayes and Mockros (1971) for confined-compression and shear studies. Subsequently, Parsons and Black (1977) extended the thin elastic-layer indentation solution of Hayes et al. (1972) to a generalized Kelvin-type viscoelastic solid. A continuous relaxation spectrum was derived from the experimental data using some approximations.

Modern mixture theories have been applied to the study of the biomechanical responses of articular cartilage (Mow et al., 1980, 1989, 1990; Mak, 1986; Mak et al., 1987; Lai et al., 1993). Mak et al. (1987) obtained a mathematical solution for the indentation creep and stress-relaxation behavior of articular cartilage. Mow et al. (1989), Spilker et al. (1992), and Suh and Spilker (1994) reported further biphasic analysis of the indentation of articular cartilage using finite element analysis. Hale et al. (1993) conducted similar finite element study using a spherical indentor. These models have also been introduced to the investigation of skin and subcutaneous soft tissues (Oomens et al., 1987a, b; Lanir et al., 1990; Mak et al., 1994b).

Fung (1972, 1981) proposed a quasi-linear viscoelastic (QLV) theory to describe the load-deformation relationship of biological soft tissues. In this theory, the load response, $T(t)$, of a tissue to an applied deformation history, $u(t)$, was expressed in terms of a convolution integral of a reduced relaxation function $G(t)$, and a nonlinear elastic function $T^e(u)$,

$$T(t) = \int_{-\infty}^{t} G(t-\xi)\cdot\frac{\partial T^e(u(\xi))}{\partial u(\xi)}\cdot\frac{\partial u(\xi)}{\partial \xi}\cdot d\xi \qquad (1)$$

The quasi-linear viscoelastic theory is one of the most successful phenomenological models for soft tissues (Johnson et al., 1996). A number of authors have used quasi-linear models to investigate the viscoelastic behavior of various biological soft tissues, such as articular cartilage (Woo et al., 1980; Simon et al., 1984; Spirt et al., 1989), passive cardiac and skeletal muscle (Pinto and Patitucci, 1980; Huyghe et al., 1991; Best et al., 1994), ligament and tendon (Lin et al., 1987; Chun and Hubbard, 1987; Woo et al., 1993; Johnson et al., 1996).

In the current paper, the linear elastic indentation solution (Hayes et al., 1972) was extended to include a QLV model. The QLV material parameters were estimated for the soft tissues around the lower limb by curve fitting the manual indentation data obtained using a hand-held ultrasound indentation probe.

Materials and Methods

Indentation Apparatus. An ultrasound indentation system for soft tissue biomechanical study was recently developed (Mak and Zheng, 1995; Zheng and Mak, 1995, 1996, 1998a, b). The indentation apparatus comprised a pen-sized, hand-held indentation probe. A cylindrical flat-ended ultrasound transducer of 9 mm diameter was located at the tip of the probe, serving also as the indentor. The thickness and deformation of the soft tissue layer were determined from the ultrasound reflection signal. A compressive load cell was connected in series with the ultrasound transducer to record the corresponding force response. Figure 1 shows the block diagram of the indentation apparatus. Figure 2 shows an in-vivo indentation test on the forearm using the pen-sized probe.

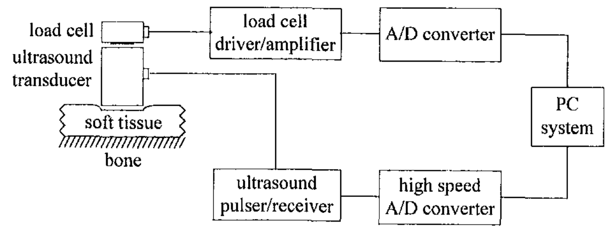

Fig. 1 Block diagram of the ultrasound indentation system

Software was developed to display in real time the ultrasound reflection signals, deformation transient, and force transient on the computer monitor. All the data could be recorded in electronic files for further off-line studies, and the whole procedure could be played back continuously or in incremental time steps. A pair of cursors within the ultrasound wave train could be set for auto-tracking the echo peaks during the indentation procedure. The ultrasound echo wave reflected from the soft tissue–bone interface moved back and forth as the soft tissue layer was loaded and unloaded. The software calculated the deflection of the time of flight of the ultrasound echo wave with the reference of the wave peak. The original tissue thickness and the subsequent change of thickness in response to loading could be derived from the time information by assuming that the sound velocity in the soft tissue was 1540 m/s.

Using a 100 MHz 8-bit A/D converter for digitizing the ultrasound signals, the accuracy of the time measurement was ± 10 ns. The accuracy of the deformation determined by the indentation system was better than 0.02 mm, with the ultrasound frequency above 5 MHz and the amplitude of ultrasound signal larger than 10 percent of the full dynamic range. A 12-bit

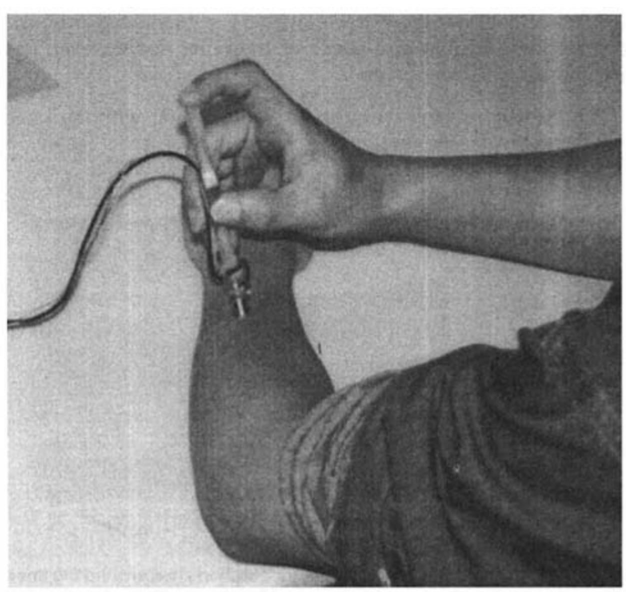

Fig. 2 Demonstration of an in-vivo test using the ultrasound indentation probe

Table 1 Some basic information on the eight subjects involved in the experiments

Information	YVO	JES	FEN	LIY	TOM	STE	TER	HJY	mean	SD
sex	F	F	F	F	M	M	M	M		
age	27	26	27	31	27	30	26	30	28.0	2.0
weight (kg)	50	51	54	47	73	65	60	60	57.5	8.7
height (cm)	152	158	156	160	165	168	183	166	163.5	9.5

A/D converter was used to digitize the force signal, the accuracy of force data was better than 0.003 N within the 10 N range. A Hounsfield material testing machine was used to validate the ultrasound indentation system (Zheng and Mak, 1996). The hand-held probe was placed in line with the load cell of the material testing machine. The applied force and the corresponding deformation of a porcine tissue specimen read from the Hounsfield machine agreed very well (variation less than 3 percent) with those determined using the ultrasound indentation apparatus. The effects of indentor misalignment were also studied and reported by Huang and Mak (1994b) and Zheng et al. (1997a).

Subjects, Testing Sites, and Body Postures. Four normal young males and four normal young females with age of 28 ± 2 years participated in the experiment. The average age of selected females was similar to that of the males. Table 1 shows some basic information on the eight subjects. Indentation tests were carried out at four different sites over the distal femur and proximal tibia and fibula.

Figure 3 shows the four indentation sites and the three postures selected for this study. The locations of the four sites at the right lower limb were:

- Site 1: On the soft tissue over the medial side of the tibia, and 3 cm from the proximal end of the tibia;
- Site 2: On the soft tissue over the lateral side of the femur, and 10 cm from the distal end of femur.
- Site 3: On the soft tissue over the lateral side of the fibula, and 5 cm from the proximal end of fibula.
- Site 4: On the soft tissue in the popliteal area at the same level as site 3.

There were three sites around the proximal tibia and fibula with different soft tissue thicknesses. Only one site was selected over the femur. These sites were chosen for their relevance to below-knee prosthetics, including those with supracondylar suspension.

The indentation tests were conducted for three body postures corresponding to different states of muscular contraction. The three postures selected were:

(A) sitting relaxed with the thigh horizontal with the knee at 90 deg flexion,

(B) posture A but with the foot dorsiflexed; and

(C) standing with the knee locked in hyperextension.

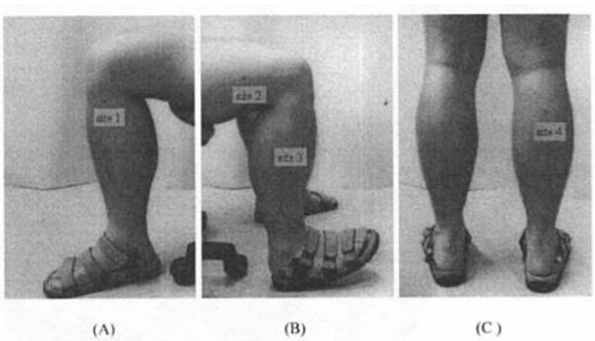

Fig. 3 The three postures and the four indentation sites around the knee of the right leg. The three postures were: (A) sitting with the knee rested at 90 degflexion, (B) posture A but with the foot dorsiflexed, and (C) standing with the knee locked in hyperextension.

The subject was asked to sit in a chair with the knee rested at 90 deg flexion, and the height of the chair was adjusted to ensure the thigh being horizontal (posture A). After the tests were completed for posture A, the subject was instructed to dorsiflex the foot as much as possible (posture B), and to maintain that position during the test. For the third posture, the subject was asked to stand with the knee locked in hyperextension without any intentional additional contraction of underlying muscles (posture C).

Sitting (posture A) and standing (posture C) are two typical human body postures even for amputees. Posture B was noted to yield a different state of muscle contraction compared to posture A with minimal change of body position and the instruction easy to follow.

Testing Protocol. Indentation test was manually performed using the pen-sized probe. Before each test, the probe was used to load and unload the testing site for several times to obtain a maximum and stable ultrasound signal reflected from the soft tissue–bone interface. Through many preliminary tests, it was found that tissue preconditioning could be achieved through the same process. Similar results have been reported in the literature (Vannah and Childress, 1988, 1996). Each test was conducted with a few cycles of loading–unloading sequences with the indentation rate manually controlled to be around 4 mm/s. Indentation rate around this value was found to be manageable for a hand-held test (Zheng and Mak, 1998a). Two hundred load-indentation data points were collected within a period of 17.2 seconds. During the tests, the maximum indentation was controlled within 30 percent of the initial thickness of the soft tissues, and the maximum indentation load was limited to 5 N. It was noted that indentation of this magnitude could be accepted by subjects without discomfort. A similar testing protocol was used by Vannah and Childress (1996). Ultrasound couplant gel was used to couple the ultrasound signal with the soft tissues.

During a test, indentation was manually controlled by the operator. The alignment of the probe was visually controlled by monitoring the amplitude of ultrasound reflection signals, which was found to be very sensitive to the misalignment of the probe (Zheng and Mak, 1997a). The indentation rate, maximum indentation, and maximum load were also visually controlled by monitoring the indentation load-deformation response shown on the computer monitor in real time. The effects of indentation rate on the results had been studied and were reported by Zheng and Mak (1997b).

Since indentation using the current apparatus was relatively simple to perform, as many as six trials were conducted at each site for each posture. Depending on the original thickness and stiffness of soft tissue at the testing site, each trial of the indentation sequence would consist of 3–9 loading–unloading cycles at an indentation rate of approximately 4 mm/s. Figure 4 shows a typical manual indentation response. The trials with the maximum and minimum responses were excluded and the mean of the remaining four data was calculated as the average result for that series of tests. The occasional movements of subjects during each trial on the results could be reduced through such an exclusion. This testing protocol and data analysis method was noted to yield repeatable results for the tests conducted on the left forearm of a normal young male subject (Zheng and Mak, 1998a). Six tests with six trials for each test were performed at a selected site. The paired t-test was used to check the statistical significance of the variations of the material parameters. Two groups of data were taken to be significantly different if $p < 0.05$.

Data Reduction. According to Hayes' (1972) indentation solution for an infinite homogeneous elastic layer on a rigid foundation:

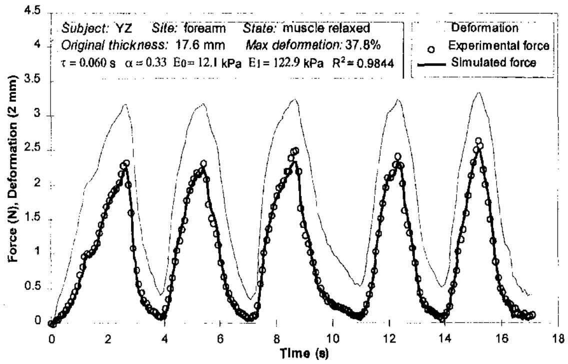

Fig. 4 A typical indentation data set and its curve-fitting result

$$P = \frac{2a\kappa(\nu, a/h)}{(1 - \nu^2)} Ew \tag{2}$$

where E is Young's modulus, w is the applied indentation, P is the indentation force, a is the radius of the indentor, h is the original thickness of the tissue, κ is a scaling factor that depends on the Poisson's ratio ν and the aspect ratio a/h. In the current study, the Poisson's ratio was taken to be 0.45 (Zheng and Mak, 1998a). The ratio of the indentation w to the original thickness h is defined as the relative indentation u, the aspect ratio is then redefined as $a/(h(1 - u))$, and the scaling factor κ becomes a function of h and u. Using these definitions, Eq. (2) can be written in the following form:

$$P(u) = \frac{2ah\kappa(h, u)}{(1 - \nu^2)} Eu \tag{3}$$

To allow for nonlinear and time-dependent properties, the instantaneous Young's modulus can be rewritten as $E(u, t)$. The Poisson's ratio is assumed to be a constant during the indentation test. When a step indentation $u_0\mathbf{1}(t)$ is applied, the force response can be represented as follows:

$$P(u_0, t) = \frac{2ah}{(1 - \nu^2)} \cdot u_0\kappa(h, u_0) \cdot E(u_0, t) \tag{4}$$

The instantaneous Young's modulus $E(u, t)$ of the soft tissue is assumed to be in the quasi-linear viscoelastic form as proposed by Fung (1981):

$$E(u, t) = E^{(e)}(u) \cdot G(t), \quad G(0) = 1 \tag{5}$$

where $G(t)$, a normalized function of time, is called the reduced relaxation function, and $E^{(e)}(u)$, assumed to be a function of u alone, is called the unrelaxed elastic modulus. In this study, $G(t)$ and $E^{(e)}(u)$ are further prescribed to be of the following forms:

$$G(t) = 1 - \alpha + \alpha e^{-t/\tau} \tag{6}$$

$$E^{(e)}(u) = E_0 + E_1 u(t) \tag{7}$$

where E_0, E_1, τ, and α are material constants. Equation (7) represents a quadratic relationship between the instantaneous indentation force and the indentation.

Substituting Eq. (5) into Eq. (4):

$$P(u_0, t) = \frac{2ah}{(1 - \nu^2)} \cdot u_0\kappa(h, u_0)E^{(e)}(u_0) \cdot G(t) \tag{8}$$

Equation (8) can be rewritten in the following form:

$$P(u_0, t) = P^{(e)}(u_0) \cdot G(t) \tag{9}$$

where

$$P^{(e)}(u) = \frac{2ah}{(1 - \nu^2)} \cdot u\kappa(h, u)E^{(e)}(u) \tag{10}$$

The function $P^{(e)}(u)$ is defined as the unrelaxed elastic indentation force response. According to the superposition principle, when a continuous indentation sequence $u(t)$ is applied, the indentation force response can be represented as follows:

$$P(t) = \int_{-\infty}^{t} G(t - \xi) \cdot \frac{\partial P^{(e)}(u(\xi))}{\partial u(\xi)} \cdot \frac{\partial u(\xi)}{\partial \xi} \cdot d\xi \tag{11}$$

With the definition $G(0) = 1$ and $u(t) = 0$ $(t < 0)$, Eq. (11) is equivalent to:

$$P(t) = P^{(e)}(u(t)) + \int_0^t P^{(e)}(u(t - \xi)) \cdot \frac{\partial G(\xi)}{\partial \xi} \cdot d\xi \tag{12}$$

Substituting Eqs. (6), (7), and (10) into Eq. (12), we find:

$$P(t) = \frac{2ah}{(1 - \nu^2)} \Bigg[\kappa(h, u(t)) \cdot [E_0 u(t) + E_1 u^2(t)] - \frac{\alpha}{\tau} \int_0^t \kappa(h, u(t - \xi)) \cdot [E_0 u1(t - \xi) + E_1 u^2(t - \xi)] \cdot e^{-\xi/\tau} \cdot d\xi \Bigg] \tag{13}$$

This equation can be rewritten in a discrete form:

$$P(i) = \frac{2ah}{(1 - \nu^2)} \Bigg[\kappa(h, u(i)) \cdot [E_0 u(i) + E_1 u^2(i)] - \frac{\alpha}{\tau} \sum_{j=1}^{i} \kappa(h, u(i - j)) \cdot [E_0 u(i - j) + E_1 u^2(i - j)] \cdot e^{-j \cdot dt/\tau} \cdot dt \Bigg] \tag{14}$$

where dt is the time interval between two contiguous data points. Using Eq. (14), the indentation force $P(i)$ can be predicted from the indentation response $u(i)$ and the four material constants E_0, E_1, τ, and α. The scaling factor $\kappa(h, u(i))$ can be obtained from the table given by Hayes et al. (1972) according to h, $u(i)$ and the assumed Poisson's ratio ν. When the indentation is small enough comparing with the original thickness or the original thickness is large enough, the scaling factor may be assumed as a constant during the whole indentation sequence. The curve-fitting procedure was completed in a VAX computer. The routine BCPOL of the IMSL mathematical library was used to search the optimal material constants yielding the minimum simulation error. The simulation error was defined as follows:

$$S_{\mathrm{err}} = \sqrt{\sum (P_S(i) - P_e(i))^2 / \sum (P_e(i))^2} \tag{15}$$

where S_{err} is the simulation error, $P_e(i)$ is the experimentally measured force, and $P_s(i)$ is the numerically simulated force. To verify the uniqueness of the simulation result using the QLV parameters, a series of curve-fitting procedures was performed for a typical indentation data set. The simulation errors were analyzed, and reported in the following section.

Results

Indentation Responses. Figure 4 shows the typical comparison of load-indentation responses between different sites

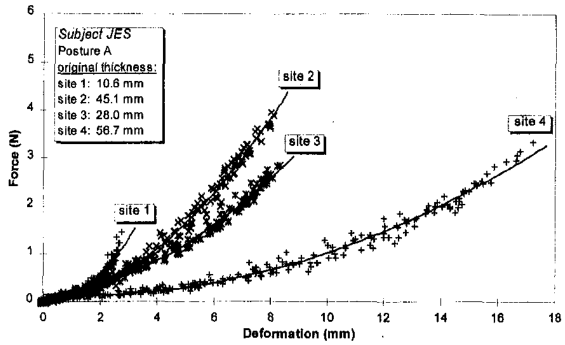

Fig. 5 Comparison of the typical load-indentation data recorded at different sites on the lower limb of subject JES in posture A. The solid lines represent the quadratic regressions of the experimental data.

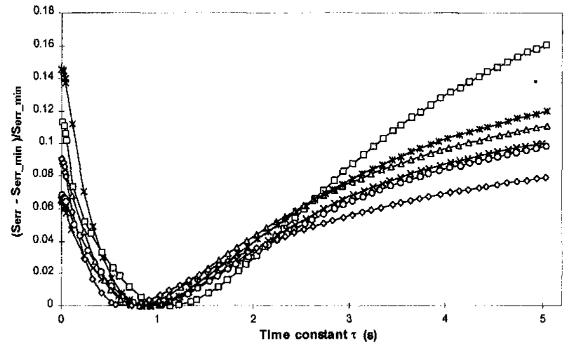

Fig. 7 Normalized simulation error versus QLV time constant obtained through curve-fitting procedures with time range changed from 0.01 ~0.02 to 4.9 ~5.0. Different symbols represent the results of the six trials of the test. The simulation error was normalized by the minimum simulation error for each trial.

recorded for subject JES. Figure 5 shows the comparison of load-indentation responses of the soft tissue over the lateral fibula (site 3) of subject JES with different postures. A quadratic form was found to be sufficient to represent most of the load-indentation responses.

Simulation Analysis. Figure 6 shows a typical indentation data set together with the curve-fitting results. The fitted curve agreed very well with the experimental data for most of the tests. Figure 7 shows the relationship between the simulation error and the simulated QLV time constant τ of the six trials with the time range changed from 0.01 ~ 0.02 s to 4.9 ~ 5.0 s. The simulation error was normalized with the minimum simulation error for each trial. The mean value of the determined QLV time constant τ with the minimum simulation error for this test was 0.92 s, and the mean percentage simulation error was 8.8 percent. It was demonstrated that the simulation result was unique for the time constant changed from 0.01 to 5.0 for this test with six trials.

Repeatability Tests. The mean values and the standard deviations of the extracted QLV parameters τ, α, E_0, and E_1 of the six groups of the repeatability test were respectively 0.052 s $\pm$ 0.003 s (6.2 percent), 0.240 $\pm$ 0.012 (4.9 percent), 13.12 kPa $\pm$0.76 (5.7 percent), 67.53 kPa $\pm$ 4.09 (6.0 percent). All the percentage standard deviations were smaller than 7 percent. The results demonstrated that the extracted QLV parameters were repeatable.

Initial Modulus E_0. The QLV model parameter E_0, that is, the initial modulus of the lower limb soft tissues was in the range of 0.22 kPa (subject STE, site 1, posture C) to 58.4 kPa (subject TOM, site 2, posture B), and the exact value depended on subjects, sites, and postures. The overall mean value was 11.2 $\pm$ 7.6 kPa. The percentage standard deviations of the eight subjects were calculated for each site and each posture. The mean of these percentage standard deviations was used to represent the subject to subject variation. The subject variation of the QLV initial modulus E_0 was 67.1 percent.

Figure 8 shows the QLV initial modulus E_0 at the four testing sites of the eight subjects in three postures. In sitting posture (posture A and B), the initial modulus at the sites over the medial proximal tibia (site 1) and the calf area (site 4) was significantly smaller than those at the other two sites. In standing posture (posture C), the initial modulus at site 1 remained small, and that at site 4 became as high as those at the other two sites. At the site over the medial proximal tibia (site 1), the initial modulus significantly decreased when the posture changed from A to B or C. At the site over the lateral femur (site 2), no significant change was observed among postures. At the sites over the lateral fibula (site 3) and the calf area (site 4), the initial modulus of the soft tissue increased significantly when the posture changed from sitting (posture A and B) to standing (posture C). The initial modulus increased when the underlying muscles were contracted.

Figure 9 shows the QLV initial modulus of male subjects was significantly larger than that of female subjects in posture A. Similar results were obtained for the other two body postures. The mean of initial moduli of the four female subjects ranged from 2.1 to 14.3 kPa, and that of the four male subjects ranged from 2.2 to 34.4 kPa. The overall mean value of the male subjects was 121 percent larger than that of the female subjects.

Nonlinear Factor E_1. The QLV nonlinear factor E_l of the lower limb soft tissues was in the range of 21.7 kPa (subject FEN, site 1, posture C) to 547 kPa (subject YVO, site 2, posture C), and the exact value depended on subjects, sites, and postures. The overall mean value was 144 $\pm$ 61 kPa. The subject

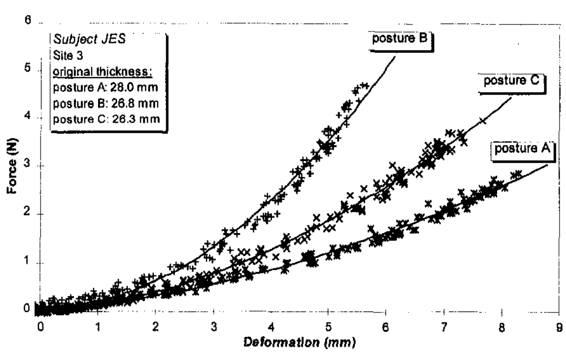

Fig. 6 Comparison of the typical load-indentation data recorded at site 3 of subject JES in different postures. The solid lines represent the quadratic regressions of the experimental data.

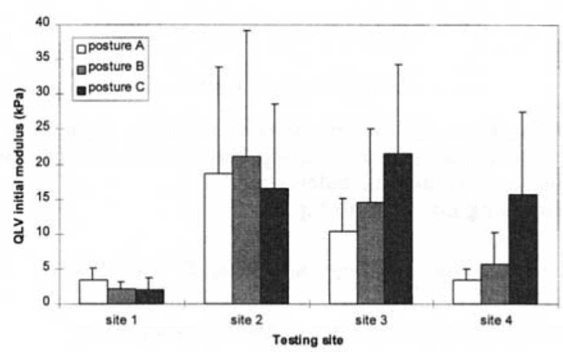

Fig. 8 Determined QLV initial modulus E_0 of the lower limb soft tissue at the four testing sites in the three body postures. The error bar represents the standard deviation of the data of the eight subjects.

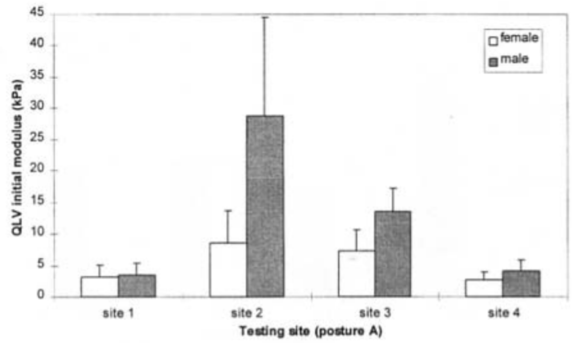

Fig. 9 Gender dependence of the QLV initial modulus E_0 of the lower limb soft tissue at the four testing sites for posture A. The error bar represents the standard deviation of the data of the four male subjects or the four female subjects.

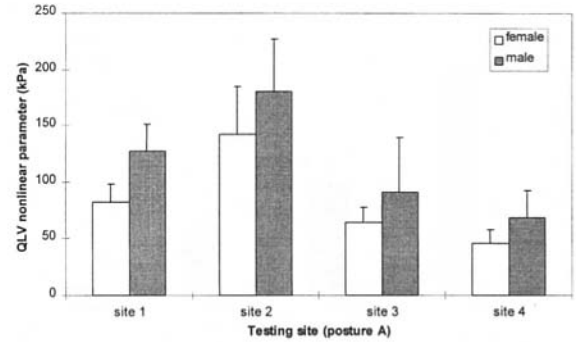

Fig. 11 Gender dependence of the QLV nonlinear factor E_1 of the lower limb soft tissue at the four testing sites for posture A. The error bar represents the standard deviation of the data of the four male subjects or the four female subjects.

variation of the QLV nonlinear factor E_1 was 40.5 percent using the same definition for the initial modulus E_0.

Figure 10 shows the QLV nonlinear factor E_1 for the four testing sites with the three postures. In the sitting posture (posture A or B), the QLV nonlinear factor at the site over the calf area (site 4) was significantly smaller than those at the other three sites. In the standing posture (posture C), it became significantly larger than those measured at the sites over the medial proximal tibia (site 1) and over the lateral fibula (site 3). At the sites over the medial proximal tibia (site 1) and the lateral fibula (site 3), the nonlinear factor in the sitting posture with the foot dorsiflexed (posture B) was larger than those in the sitting posture with the foot relaxed (posture A) or those standing (posture C). The nonlinear factor increased when the underlying muscles were contracted.

Figure 11 shows the QLV nonlinear factor E_1 of male subjects was significantly larger than that of female subjects in posture A. Similar results were obtained for the other two body postures. The nonlinear factors of the four female subjects ranged from 46.4 to 204.8 kPa, and those of the four male subjects ranged from 69.2 to 250.3 kPa. The overall mean value of the male subjects was 41.8 percent larger than that of the female subjects.

Time Constant τ. The QLV time constant τ of the lower limb soft tissues was in the range of 0.05 s (subject FEN, site 1, posture A) to 8.93 s (subject LIY, site 3, posture C), and the exact value depended on subjects, sites, and postures. The overall mean value was 4.69 s $\pm$ 1.79 s. The subject variation of the QLV time constant τ was 64.3 percent using the same definition of subject variation for the initial modulus E_0.

Figure 12 shows the QLV time constant τ for the four testing sites with the three postures. In posture A, the time constant of soft tissue at the calf area (site 4) was significantly smaller than those at the other three sites. In posture B, those measured at sites 2 and 3 were significantly larger than those at sites 1 and 4. In posture C, no significant difference was found among the four sites. At the sites over the lateral femur (site 2), the lateral fibula (site 3) and the calf area (site 4), the time constant of the soft tissue increased when the posture changed from sitting with the foot relaxed (posture A) to sitting with the foot dorsiflexed (posture B) or standing (posture C). It was demonstrated that the time constant increased when the underlying muscles were contracted.

Figure 13 shows the QLV time constant τ significantly depended on gender for the four testing sites with posture A. Similar results were obtained for the other two body postures. However, the overall pattern between the time constant and gender could not be established from the result. The time constants of the four female subjects ranged from 0.96 to 7.95 s, and those of the four male subjects ranged from 0.12 to 7.08 s.

Parameter α. The QLV parameter α of the lower limb soft tissues was in the range of 0.029 (subject TOM, site 4, posture C) to 0.277 (subject HJY, site 1, posture B), and again its exact value depended on subjects, sites, and postures. The overall mean value was 0.132 $\pm$ 0.041. The subject variation of the QLV parameter α was 40.1 percent using the same definition of subject variation for the initial modulus E_0.

Figure 14 shows the determined QLV parameter α for the four testing sites with the three postures. In all the three postures, the parameter α of soft tissue at the site over the medial proximal tibia (site 1) was significantly larger than those at the other three sites. No significant change in the parameter α could be ascribed to the change of the postures at all the four sites.

Figure 15 shows the relationship between the QLV parameter α and the gender for the four testing sites with posture A.

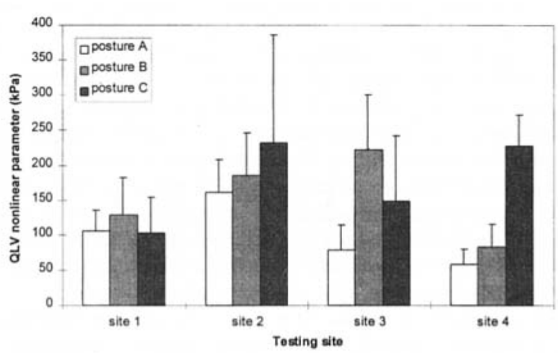

Fig. 10 QLV nonlinear factor E_1 of the lower limb soft tissue determined at the four testing sites for the three body postures. The error bar represents the standard deviation of the data of the eight subjects.

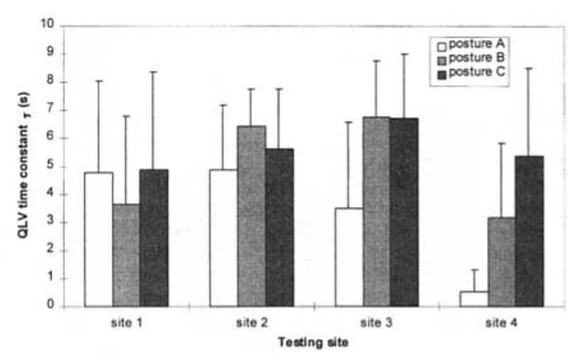

Fig. 12 QLV time constant τ of the lower limb soft tissue determined at the four testing sites for the three body postures. The error bar represents the standard deviation of the data of the eight subjects.

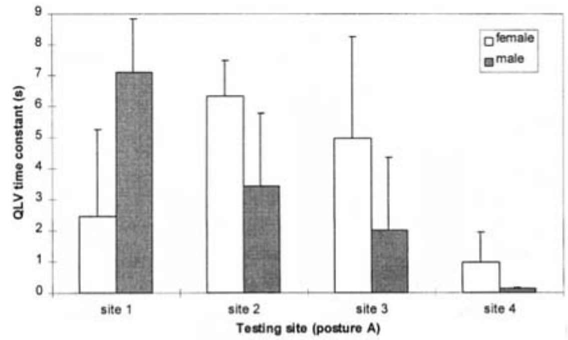

Fig. 13 Gender dependence of the QLV time constant τ of the lower limb soft tissue at the four testing sites for posture A. The error bar represents the standard deviation of the data of the four male subjects or the four female subjects.

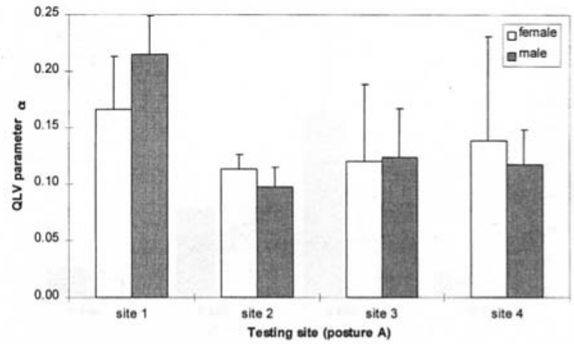

Fig. 15 Gender dependence of the QLV parameter α of the lower limb soft tissue at the four testing sites for posture A. The error bar represents the standard deviation of the data of the four male subjects or the four female subjects.

There was no significant dependence of the determined QLV parameter α on gender. Similar results were obtained for the other two body postures. The parameter α of the four female subjects ranged from 0.098 to 0.197, and that of the four male subjects ranged from 0.077 to 0.215.

Discussion

In this paper, a linear elastic indentation solution was extended to include a quasi-linear viscoelastic (QLV) model to extract the nonlinear and time-dependent properties of limb soft tissues. The QLV parameters were obtained by curve fitting the experimental indentation data. The indentation responses were collected using an ultrasound indentation apparatus with a hand-held pen-sized probe.

In the literature, there were few reports on the nonlinear or viscoelastic properties of limb soft tissues. Most of the earlier investigators only reported an effective Young's modulus assuming the soft tissue as linear elastic. The moduli so determined reflected both material and geometric factors and thus could not be taken as genuine material properties. The results of the extracted moduli of the lower limb soft tissues in the literature were: 60 kPa (Steege et al., 1987), 53 kPa to 141 kPa (Krouskop et al., 1987), 50 kPa to 145 kPa (Reynolds and Lord, 1988, 1992), 27 kPa to 106 kPa (Torres-Moreno et al., 1991, 1992b), 21 kPa to 194 kPa (Mak et al., 1994a), 2 to 600 kPa (Vannah and Childress, 1996), 10 kPa to 89 kPa (Zheng and Mak, 1998a). These values varied among subjects, with sites and with states of the muscular contraction. Using the average values of the initial modulus and the nonlinear parameter obtained in this research, the modulus was calculated to be 11.2 kPa, 25.6 kPa, 40.0 kPa and 54.4 kPa for percentage deformation of 0 percent, 10 percent, 20 percent, and 30 percent of the original thickness, respectively. It increased about 5 times from the initial state to the state of 30 percent deformation. The effective Young's modulus obtained by Vannah and Childress (1996) at a narrowly constrained site in the calf area using a strain energy function increased from 2 to 600 kPa as the applied load changed from 0 N to 7 N with a 8 mm diameter indentor. The effective Young's modulus of soft tissues at the similar area calculated from the QLV initial modulus and nonlinear factor determined in this study increased from 3.3 to 223 kPa (Zheng, 1997), as the applied load changed from 0 N to 5 N with a 9-mm-dia indentor. Different experimental conditions might be the reason that the maximum modulus determined in this study was smaller than that obtained by Vannah and Childress (1996). The dependence of the effective Young's modulus on the applied load suggested that numerical analyses of residual limbs should take into account of the nonlinear properties of the soft tissue. This is particularly relevant to the analyses of stresses built up in the residual limbs during the cloning process.

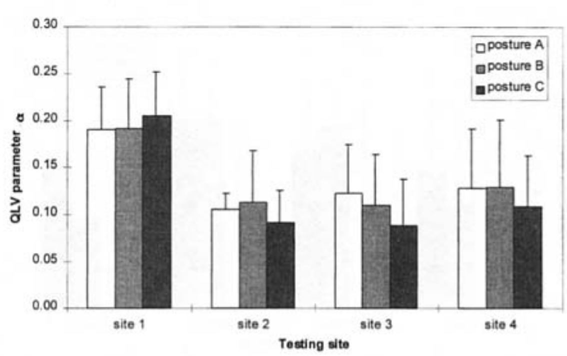

Fig. 14 QLV parameter α of the lower limb soft tissue determined at the four testing sites for the three body postures. The error bar represents the standard deviation of the data of the eight subjects.

The repeatability tests on a forearm site showed that the determined QLV parameters were repeatable using the reported testing protocol. The initial modulus and nonlinear factor significantly increased when the underlying muscles were contracted. These results agreed with the fact that the limb tissues stiffen when its muscles were contracted. The initial modulus and nonlinear factor also depended significantly on the testing site, subject, and gender. These parameters tended to be larger for the male subjects compared to the female. It was also demonstrated that the QLV time constant τ significantly increased as the underlying muscles were contracted. Although an overall pattern between the gender and the QLV time constant of limb soft tissues could not be established from the results of this study, there existed significant difference of the data between the male and the female subjects for individual sites and postures. The biomechanical and biological reasons for the behaviors of the QLV time constant deserved to be further investigated. Another QLV parameter α reflected the viscous behavior of the soft tissues tested. The parameter α depended significantly on subjects and sites. Its dependence on postures or genders was, however, not demonstrated in this study. Further studies could be conducted on the correlation between the QLV parameters and the age or pathological conditions of limb soft tissues. Compared to the palpation method being used in clinics or the earlier approaches for determining the effective Young's modulus, the QLV parameters obtained in this study represented a significant advance in limb tissues characterization.

Large subject variations were observed in the determined QLV parameters. Besides the genuine subject dependence of these material parameters, another main source for these variations might be the uncertainty of the contraction state of the underlying muscles among subjects for a specific testing site in a given posture. It has been demonstrated that most of the determined QLV parameters were very sensitive to the muscular contraction state. In the current testing protocol, the subjects were instructed to maintain a specific body posture during a test. However, there was no quantitative criterion used to monitor the degree of muscle contraction. It was possible that different subjects might involve different degrees of muscle contraction to maintain a specific posture, especially for posture B (sitting with foot dorsiflexed) and posture C (standing). This might be a main source for the apparent subject variations of these QLV parameters.

In most indentation studies on skin and subcutaneous tissues or FE modeling of limb soft tissues, investigators have assigned a Poisson's ratio within the range from 0.45 to 0.5 to simulate the nearly incompressible behavior of the limb soft tissues as a whole. The Poisson's ratio ν was selected to be 0.45 in this study. Our previous study on the extraction of the effective Young's modulus demonstrated that the uncertainty of the Poisson's ratio of soft tissue would affect the results (Zheng and Mak, 1998b; Zheng, 1997). If the Poisson's ratio is assumed to be 0.3 instead of 0.45 for the indentor diameter of 9 mm used in the study, the extracted effective Young's modulus will increase by 14.1 percent to 19.4 percent as the tissue thickness changes from infinite to 20 mm. However, if the Poisson's ratio is assumed to be 0.5 instead of 0.45, the extracted effective Young's modulus will decrease by 6.0 percent to 8.6 percent as the tissue thickness changes from infinite to 20 mm. This effect would become more serious when the tissue thickness decreased relative to the size of the indentor. The uncertainty of the Poisson's ratio of soft tissue would also affect the QLV parameters determined in this paper. It is likely that the Poisson's ratio might be different for various subjects, sites, and states of muscular activity. The Poisson's ratio of limb tissues in vivo deserves to be investigated further.

Besides the Poisson's ratio, the geometry of the limb soft tissues might be another source of uncertainty for the determined QLV parameters. The indentation solution (Eq. (2)) employed in this study was derived for a infinite uniform layer. Significant curvature of limb surface and the nonuniform tissue thickness in the indented region might affect the accuracy of the material parameters determined. Selection of a smaller indentor could reduce such effects. However, too small an indentor may pick up other variations due to skin hairs and other epidermal surface characteristics. Numerical and experimental studies are needed to examine the effects of the curvature of the limb surface as well as the size and shape of underlying bone.

The QLV parameters were determined by a least-square curve-fitting procedure in this study. Since there were four parameters used in the simulation, the uniqueness of the parameters was a concern. In this study, the curve-fitting error was analyzed for the simulated QLV time constant τ of the six trials of a test with the time range from 0.01 to 5.0 s. The procedure was equivalent to decrease the number of the parameters. It demonstrated that the curve-fitting error did show a minimum for the time constant from 0.01 to 5.0 for the test with six trials.

In spite of these concerns, the nonlinear and viscoelastic properties of the soft tissues determined in this study could provide more information about the condition of the soft tissues than a single effective Young's modulus. With the easy-to-use indentation apparatus, the approach introduced in this paper could be used to evaluate the biomechanical integrity of biological soft tissues, such as in the assessment of residual limb soft tissues. The QLV parameters of limb soft tissues would also be useful in the computational analysis of prosthetic mechanics.

It should be noted that the parameters were calculated based on an ad-hoc extension of the isotropic linearly elastic layer solution. For isotropic material, which the limb soft tissues were assumed to be in this case, it is possible to convert the QLV initial modulus and nonlinear factor to the parameters of strain-energy functions as employed by Vannah and Childress (1996). A three-dimensional QLV model has been proposed by Fung (1981), and a computational model of residual limb including these QLV constants could be further investigated. QLV parameters determined in this study are the phenomenological material properties of soft tissues. Studies in using other structural biomechanical properties can be further explored. The biphasic model has been successfully applied to the indentation of articular cartilage (Mak et al., 1987; Mow et al., 1989; Spilker et al., 1992; Suh and Spilker, 1994). A preliminary study demonstrated that a linear biphasic model was not adequate to represent the load-indentation responses of limb soft tissues (Liu, 1993). Further investigations are being carried out using nonlinear biphasic models.

Acknowledgments

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