An Ultrasound Indentation System for Biomechanical Properties Assessment of Soft Tissues *In-Vivo*

Yong-Ping Zheng, *Student Member, IEEE*, and Arthur F. T. Mak*

Abstract—**An ultrasound indentation system for biomechanical assessment of soft tissues *in vivo* was developed. The pen-size, hand-held probe was composed of an ultrasound transducer and a load cell. The ultrasound transducer was 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 echo. A compressive load cell was connected in series with the ultrasound transducer to record the force response. A validation experiment was performed on porcine tissues. Force and deformation acquired with the present system was in good comparison with those obtained from a Housfield material testing machine. Material constants were obtained via a curve-fitting procedure by predicting the force transient response from the deformation-time data using a quasilinear viscoelastic model. In addition, deformation in the fat and in the muscle could be differentiated. The potential applications of this type of indentation probes are many. The specific application of this current development is for stump tissue assessment in the design of prosthetics.**

I. Introduction

BIOMECHANICAL properties of tissues are known to vary from site to site and change with age and pathological conditions. These properties are known to be time-dependent, i.e., viscoelastic, and direction-dependent, i.e., anisotropic. Organizationally, biological tissues are arranged in various hierarchy to form structures and organs. Their properties can be different depending on whether they are measured *in-vivo* or *in-vitro*, *in situ* or as an excised specimen. For the lack of quantitative biomechanical tools for *in-vivo* assessment, palpation is still widely used to evaluate firmness of soft tissues in common clinical practice. The aim of our study is to develop a clinical apparatus to assess the biomechanical properties of soft tissue on bony substratum.

Several generations of *in-vivo* soft tissue indentation assessment devices have been developed for various applications over the years [1]–[8]. Schade first developed an apparatus for the study of edema [1]. Kirk *et al.* utilized a similar approach to compare the elastic properties of skin and subcutaneous tissues in young and old individuals [2]. Ziegert *et al.* performed indentation creep experiments on the thin layer of soft tissues covering the anterior medial aspect of human tibia [3]. Lewis *et al.* developed a subcutaneous fat thickness caliper device to assess subcutaneous edema [4]. Vannah *et al.* designed an indentation device with a holding shell to conduct indentation tests on the calf area of human subjects [5]. Recently, a computer controlled indentation apparatus was proposed by Mak *et al.* for the study of biomechanical properties of the below knee stump tissues and of the tissues of normal young adults [6]. A pneumatic indentation system using a copper bellows was developed by Ferguson-Pell *et al.* to study the physiological responses associated with pressure sores [7]. Horikawa *et al.* developed a portable equipment to study the hardness of muscular tissues using a laser distance monitor to determine the displacement of the indentor [8]. Although the technique required the tissue surface to be relatively flat, this was an attempt to design a relatively compact and portable system for soft tissue indention.

The above indentation devices measured the epidermal displacement in concert with the displacement of the indentor [1]–[9]. The thickness of the tissue was not monitored. The stiffness measured in this manner would reflect not only the material properties of the tissues but also some geometric effects. In the present assessment methodology, ultrasound technique was employed to determine the change of the thickness of the soft tissue layer in the indented site as a function of time. With this approach and with the help of some theoretical models, one could separate the geometric effects from the indentation data and extract from the data the genuine material properties of the tissue layers.

Diagnostic ultrasound techniques in terms of B-scan backscatter imaging and Doppler blood flow measurement in vessels and heart have proven to be successful biomedical applications of the ultrasound technology. Ultrasound was also used to monitor the internal displacement such as that of heart wall and fetal lung [10]. Recently, much research has been directed to the imaging of strain and tissue elasticity [11]–[21]. There are mainly two kinds of techniques to obtain those images, based on the two traditional diagnostic applications of ultrasound, respectively. One method imposed an external source of forced vibration and used the Doppler technique to determine the internal vibration of soft tissue and to generate the relevant image [11]–[14]. This vibration technique was proposed by Krouskop *et al.* for application in stump assessment [11]. A single ultrasound transducer was employed in their study, and the soft tissue on the lateral side of an above-knee amputee's residual limb was tested. Another method consisted of the application of an external indentor, usually the ultrasound transducer itself. Standard B-scan

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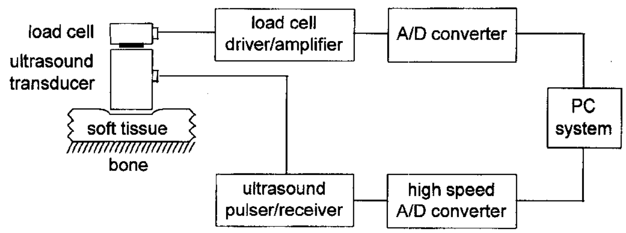

Fig. 1. A schematic showing the entire ultrasound indentation system.

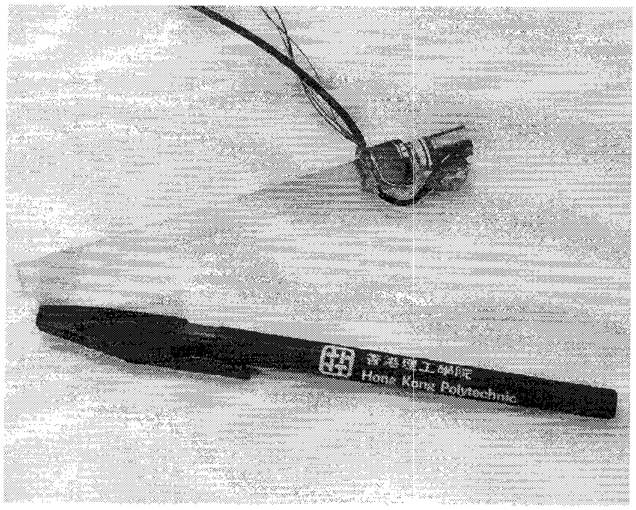

Fig. 2. The ultrasound indentation probe.

images and the corresponding ultrasonic radio frequency (RF) signals were recorded before and after the force was applied. A segmental cross-correlation procedure was employed to determine the deflection of the echo train to derive the strain or elasticity images [15]–[21]. In pathological situations such as cancer, edema, degeneration, and pressure sore, there are usually changes in the soft tissue biomechanical properties. These changes are sometimes difficult to detect with the normal B-scans. Thus, the new elastography type of imaging has interesting potential [21]. However, some limitations are still associated with the current state-of-the-art. For example, usually only small deformation could be applied and the soft tissues were assumed to be linearly elastic. Furthermore, the lack of precise experimental stress field measurement would limit most applications to situations where the assumption of a homogeneous stress field could be justified [21].

In the literature, a number of models have been presented to analyze the indentation experiment on soft tissue [22]–[27]. Hayes *et al.* described a mathematical analysis of indentation using a linear elastic model for a homogeneous tissue layer such as articular cartilage [22]. This was extended later by Parsons and Black to include linear viscoelasticity [23]. Mak *et al.* re-examined the indentation analysis using a biphasic model, assuming the tissue layer to be a hydrated poroelastic structure [24]. Mak and Liu applied a quasilinear viscoelastic theory to analyze the experimental indentation data collected on the soft tissues around human proximal tibiae, both normal and of below-knee amputees [6], [25].

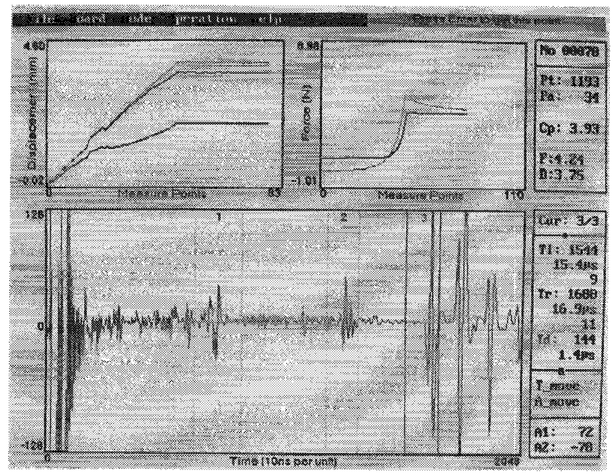

Fig. 3. The user interface of the software. The lower window showed the ultrasound echo train and the auto-tracking cursors. The upper left window showed the deformation-time curve. The upper right window showed the force-time curve. The period between two contiguous data points in the top windows was 0.08 s.

In the present paper, an autotracking method was developed to track the ultrasound echo train to obtain continuous measurement of tissue deformation [28]. A quasilinear viscoelastic model was employed to perform curve fitting to extract the material constants from the data. Our aim was to develop an easy-to-use indentation probe and a simple protocol for assessing the biomechanical properties of soft tissue layers *in-vivo*.

II. Ultrasound Indentation System

The schematic diagram of the present ultrasound indentation system is shown in Fig. 1. The ultrasound transducer was used not only to transmit and receive ultrasound waves, but also to serve as the indentor. Several Panametrics ultrasound transducers of 0.25-inch dia./5 MHz, 0.25-inch dia./10 MHz, and 0.175 inch dia./10 MHz could be connected to our probe. The ultrasound transducer of 0.25-inch dia./5 MHz was used in the experiments described in following sections. A low-profile Entran compressive load cell calibrated for a 10-N range was connected in series with the ultrasound transducer via a plexiglass "plug-in" socket, which made it easy to change ultrasound transducers of different size and frequency. A perspex rod was attached to the other surface of the load cell for hand-holding. The pen-size, hand-held probe is shown in Fig. 2. A load cell driver/amplifier (model PS-30A-2 from Entran) was used to drive the load cell, and the output signal was digitized by a 12-b analogue–digital (AD) converter. A Panametrics model 5052UA ultrasound pulser/receiver was utilized to drive the ultrasound transducer and to amplify the received signal. A Sonix 100-MHz high-speed 8-b AD converter was employed to digitize the amplified ultrasound echo train.

A software was developed in Microsoft C and its user-interface is shown in Fig. 3. During the indentation process, the ultrasound signal was shown in real time in the main window at the bottom of the screen, the two windows at the

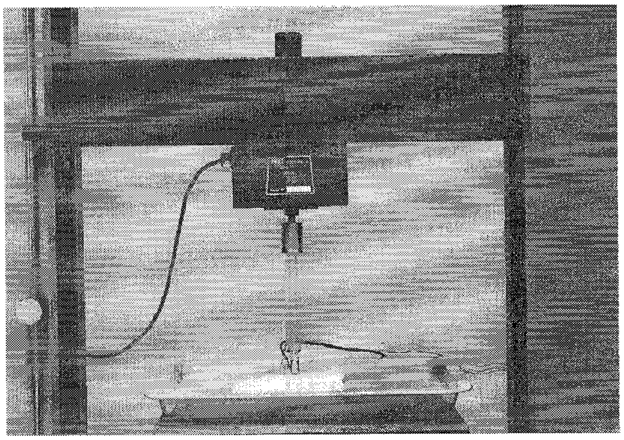

Fig. 4. A validation experiment being carried out in a Hounsfield material testing machine. The probe was placed in line with the load cell of the machine.

top showed the deformation transient, the force transient or force/deformation curve. The minimum period between two contiguous data points in the top windows was 0.08 s in the current setting. All the data could be recorded in a file for further off-line study, the whole procedure could be played back continuously or in incremental time step. More than one 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 will move back or forward when the soft tissue layer is indented or recovered. The software calculated the deflection of the time of flight of the ultrasound echo wave with the reference of the wave peak. The pair of cursors would auto-track the moving echo peak during tissue indentation and recovery. The original tissue thickness and the subsequent change of thickness, i.e., deformation of the tissue, could be derived from the time information.

With the 100-MHz 8-b AD converter, 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% of the full dynamic range. A 12-b AD converter was used to digitize the force signal, the accuracy of force result was better than 0.003 Newton within the 10-Newton range.

III. System Validation

We used a Hounsfield material testing machine to validate the ultrasound indentation system. The hand-held probe was placed in line with the load cell of the Housfield material testing machine. Fig. 4 shows the apparatus. The experiment was performed on a layer of porcine tissue with initial thickness of about 24 mm. An ultrasound velocity of 1540 m/s was assumed for the tissues. The Hounsfield material testing machine was controlled by a special PC-based software developed in-house. Force and deformation data were read from the machine by the software via the RS232 port. Similar data were also acquired with the ultrasound indentation system. One set of the validation results is shown in Fig. 5. The whole procedure consisted of several indentation-holding-recovery phases imposed via the Hounsfield control panel. The rate imposed varied from 0.5 mm/s to 1 mm/s. The largest force was about 5 N corresponding to about 20% strain of the entire layer. It was found that data from the Hounsfield machine and the new indentation system agreed very well.

IV. Extraction of Material Parameters

A number of *in-vitro* experiments were done on porcine tissues using this ultrasound indentation system. The tissue layer had dimensions of 10 cm $\times$ 10 cm $\times$ 2.5 cm. The tissues were kept moist by saline drippings throughout the experiments. The ultrasound signal was reflected from the interface between the soft tissue layer and the surface of the container. Fig. 6 shows the force-deformation curve of a typical indentation-holding-recovery procedure. The biomechanical response obviously demonstrated nonlinear and time-dependent viscoelastic behaviors.

A quasilinear viscoelastic (QLV) model of soft tissue was used to extract the relevant material parameter from the experimental data [26]. According to the model, the force response can be derived from

$$T(t) = T^{(e)}[\lambda(t)] + \int_0^t T^{(e)}[\lambda(t-\tau)]\frac{\partial G(\tau)}{\partial \tau}d\tau \quad G(0) = 1 \tag{1}$$

where, $T(t)$ is the force response at time t, $T^{(e)}(\lambda)$ is called the elastic response, $G(t)$, a normalized function of time, is called the reduced relaxation function, and $\lambda(t)$ is the deformation history. In our current study, the elastic response was taken to be a third-order polynomial function, and the relaxation function was assumed to be a single exponential function as follows:

$$\begin{aligned} G(t) &= 1 - a(1-\exp(-t/t_e)) \\ T^{(e)}(\lambda) &= C_1\lambda + C_2\lambda^2 + C_3\lambda^3 \end{aligned} \tag{2}$$

where, $a, t_e, C_1, C_2,$ and C_3 represent the five parameters of the model. Fig. 7 shows the curve-fitting result of the force response. Reasonable good agreement between the experiment data and the model simulation was obtained. The set of parameters for this particular tissue layer was determined to be

$$a = 0.5620, \quad t_e = 1.6483 \text{ (s)}, \quad C_1 = 0.1946 \text{ (N/mm)}$$
$$C_2 = 0.0213 \text{ (N/mm}^2) \quad C_3 = 0.0125 \text{ (N/mm}^3).$$

V. Deformations in Multilayered Tissue

To demonstrate how the same indentation probe might be used to quantify deformations in different layers of a tissue, two cursor windows were set to track two ultrasound echoes, one from the bottom surface of the porcine tissue specimen, and the other from an intermediate muscle-fat interface. Using this approach, the internal deformation in the muscle layer and the fat layer as functions of time could be monitored in real time. The movement of the indentor was prescribed in the similar pattern as in Fig. 6. The results are shown in Fig. 8. The percentage deformation was represented in deformation/thickness (%). At the start of indentation, the

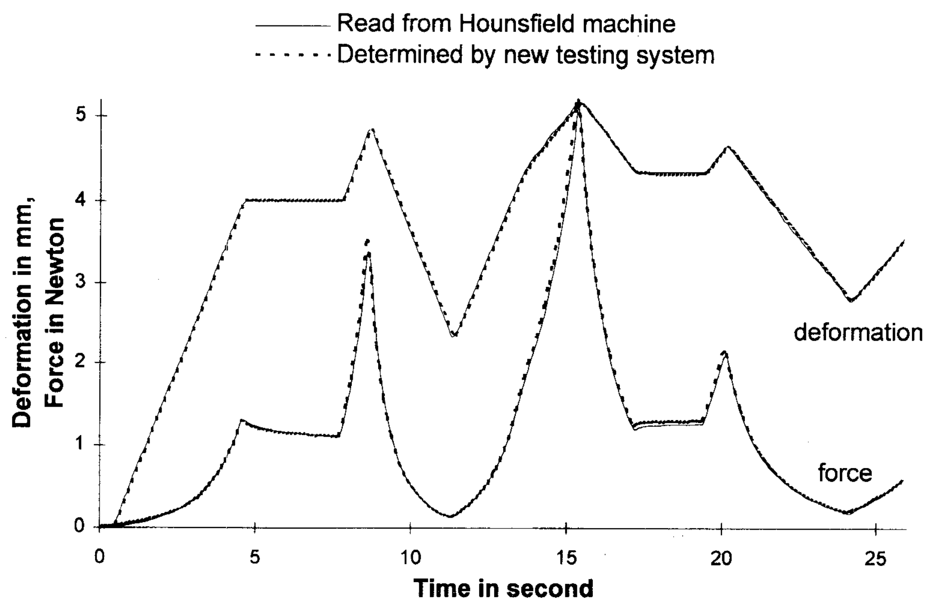

Fig. 5. Comparison of the force and deformation data measured by the Hounsfield machine and those measured using the ultrasound indentation probe. The test was carried on a fresh porcine tissue with thickness of 24.1 mm. Indentation rate was controlled within 0.5 $\sim$ 1 mm/s.

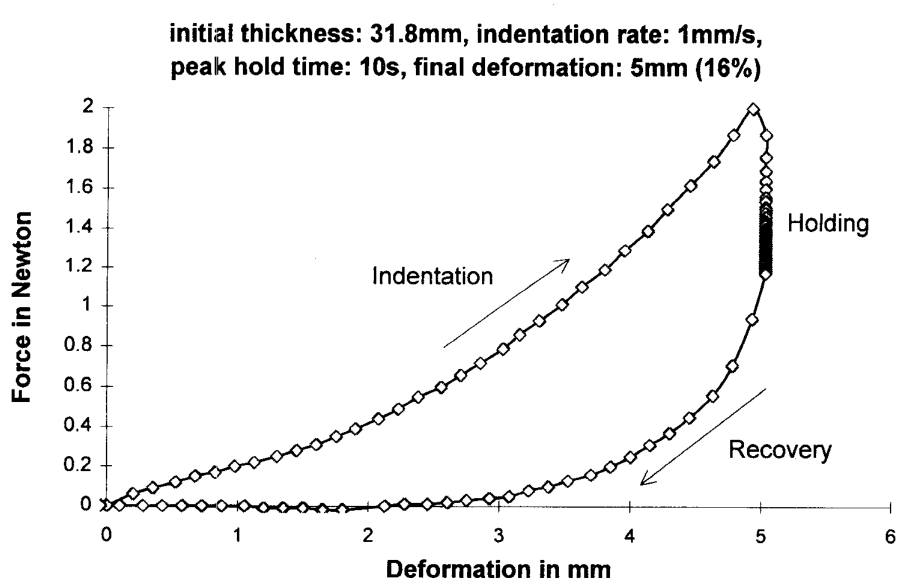

Fig. 6. A typical force-deformation curve recorded by the ultrasound indentation probe on porcine tissue. The whole procedure was composed of three phases: Indentation, holding, and recovery.

percentage deformation of the fat layer increased slowly. The rate of change of the percentage deformation later increased as indentation increased. The percentage deformation of the muscle layer was higher than that of fat. The rate of change of the percentage deformation, however, decreased as indentation increased. At the end of the indentation process, the fat layer did not fully recover and a residual percentage deformation about 4% of its initial thickness still remained. The fat layer seemed to take longer to recover. The resistive force from the soft tissue decreased to zero before the indentor fully returned to its starting position. Although the tissue surface might not follow the indenting ultrasound transducer during the later part of the recovery phase, the ultrasound echo train from the internal soft tissue could still be obtained, since the gap between the transducer and tissue surface was filled by a fluid film of saline solution. The ability of the probe to pick up nonhomogeneous time-dependent deformation in different sublayers within a tissue could be useful when more precise biomechanical assessments are required for specific tissue sublayers.

VI. Discussion and Further Studies

In vivo indentation testing is a useful method to determine the biomechanical properties of soft tissue. This paper

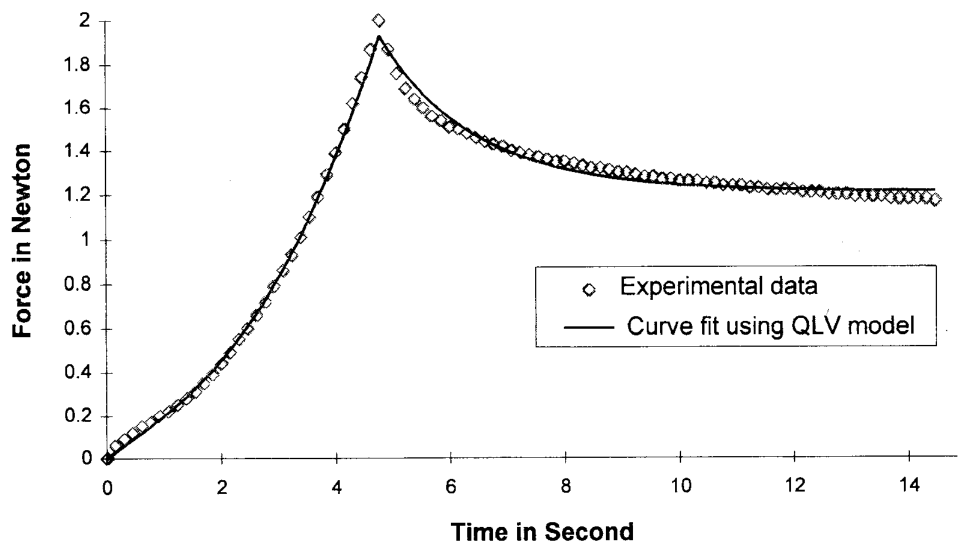

Fig. 7. Curve-fitting of the experimental force data using a quasilinear viscoelastic model. The test was carried out on a porcine tissue layer with thickness of 31.8 mm. Final deformation was 5 mm. The mark represents the experimental data, and the solid line indicates the curve-fitting result.

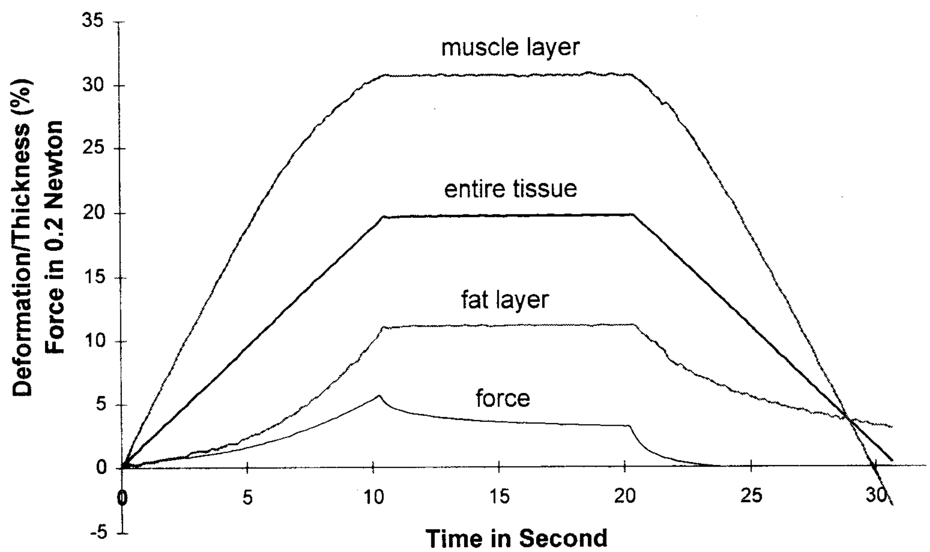

Fig. 8. The use of the ultrasound indentation probe to determine deformation in different layers of the porcine tissue. Axial strain was calculated as deformation per original thickness of the layer. The original thickness of entire tissue layer was 25.9 mm, that of the muscle layer was 11.3 mm, and that of the fat layer was 14.6 mm. The probe was driven by the Hounsfield machine in a rate of 0.5 mm/s. The final deformation of the entire tissue layer was about 20%.

described a newly developed ultrasound indentation system using a pen-size indentation probe. Validation of the ultrasound indentation system was performed on porcine tissue.

To demonstrate the feasibility of using this probe in a hand-held manner to collect *in-vivo* indentation data on human subjects, an indentation experiment was performed on the soft tissues covering the volar side of the fist metacarpal of a human subject. The probe was hand-held and visually aligned normal to the skin surface. A few cycles of indentation were imposed manually. Ultrasound signal was reflected from the soft tissue-bone interface. Fig. 9 shows a typical set of force and deformation transients.

In the present paper, simulation using a QLV model was performed to fit the *in-vitro* indentation data on the fresh porcine tissue. Good comparison was found between the model and the experimental data. In the *in-vivo* indentation experiment, since the full deformation-time record was known from the ultrasound data, the force response could also be curve-fitted using QLV model to obtain the material parameters.

Multilayer study seems possible using this ultrasound indentation system. In our preliminary study, tissues with two layers, namely a fat and a muscle layer, were examined. In principle, more than two layers can also be examined using the ultrasound indentation system by setting more tracking cursor windows.

In the present indentation apparatus, the time of flight of ultrasound wave was used to determine the deformation of the soft tissue. The amplitude of the echo train was not so critical

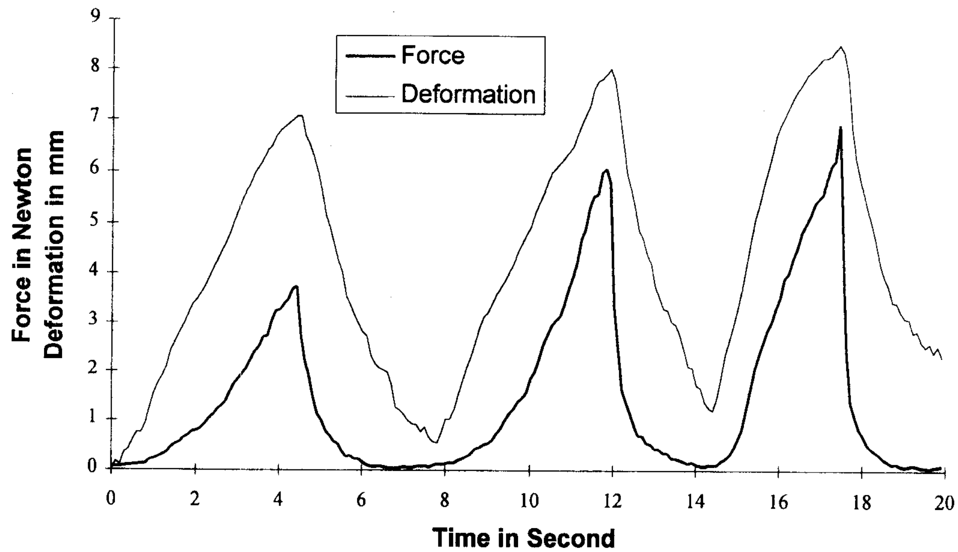

Fig. 9. Preliminary *in-vivo* result from the indentation test on the soft tissue covering the volar side of first metacarpal. The probe was driven manually by hand.

in our application so long as it was within an acceptable signal to noise ratio. The larger the width of the ultrasound beam, the longer the echo train from an interface would be. To overcome this, one could set the tracking window in the beginning of such echo train. In addition, if the internal interface which caused reflection of ultrasound wave was not flat, the reflected signal could only represent some average position of that interface [29].

If the probe is to be used in a hand-held manner, the problem of probe alignment should also be addressed. Theoretical finite element analysis showed that while probe misalignment might affect the stress distribution beneath the indentor, the effect on the total force acting on the indentor seemed to be little for misalignment up to 5° [30].

It was also expected that tissue layers with bony interface not parallel to the skin surface may compromise the amount of ultrasound reflection that could be picked up by the transducer. These issues would be examined in future reports.

The methodologies developed in the present project could be also utilized in the imaging of strain or elasticity of other soft tissues.

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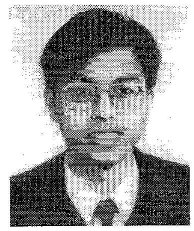

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