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Portable Gage for Pressure Ulcer Detection

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***Abstract*— Pressure ulcers are widely considered to be a critical problem in rehabilitation since they result in severe discomfort and high healthcare cost. The prevention of pressure ulcers is a constant preoccupation for every nursing team. This paper introduces a novel handheld instrument that can detect subtle changes in the skin biomechanical properties by measuring its biomechanical response. This could be used to detect stage-I pressure ulcers and deep tissue injury. Its high bandwidth makes it possible to load the skin under wide range of conditions. The instrument is portable, inexpensive, and intrinsically precise. Several experiments were conducted to validate the function of the device. Preliminary results show that the device could effectively measure the difference in the viscoelasticity between human skin of different sites, hence paving the way for the development of clinical protocols and trials.**

I. INTRODUCTION

Pressure ulcers are a significant secondary complication of mobility impairment. More than 50% of individuals with spinal cord injury (SCI) will develop a pressure ulcer during their lifetime, and the annual medicare cost for pressure ulcers in the US is around $1.3 billion, which accounts for 25% of the total health care cost for SCI [1]. They also affect older people and persons with other disabilities such as strokes, amputations and dementia. In addition to monetary cost, the non-monetary costs such as discomfort, stress, and decreased quality of life are costly to patients. The prevention of pressure ulcers is proposed to be "the principal aim for every member of the nursing home team, including residents" [2]. Early detection of pressure ulcers is not always a simple matter. The prevalent preventative strategies for pressure ulcers are comprehensive risk assessment scales such as Braden Scale and clinical inspection. The subjectivity of comprehensive risk assessment and some clinical assessments, such as visual inspection and palpation, makes the accuracy of the assessment dependent on expert experience. These limitations make these techniques more like a tendency predictor rather than a risk predictor [3].

Many methods have been explored to objectively assess the pressure ulcer risks. In [4], a bioimpedance spectrometer was proposed to detect early pressure ulcers. Other approaches used color images to analyze the presence of skin erythema [5]. Generally speaking, the detection of a stage-I ulcer is critical because the skin is still intact and it is easier to recover from this condition. According to the National Pressure Ulcer Advisory Panel (NPUAP), a stage-I ulcer is defined to be "*an observable pressure related alteration of intact skin whose indicators as compared to an adjacent or opposite area on the body may include changes in one or more of the following: skin temperature (warmth or coolness), tissue consistency (firm or boggy feel) and/or sensation (pain, itching)*". An important symptom of stage-I pressure ulcer is the change of tissue consistency. Since the skin and subcutaneous tissues start degrading on the early stage of the pressure ulcers [6], it is likely that the biomechanical properties of the skin begin to change simultaneously. The changes can be detected in a number of ways. Significant differences in material properties, such as relaxation time and elastic response, of buttock soft tissue between healthy individuals and individuals with pressure ulcer susceptibility have been observed [7]. Therefore, the risk of pressure ulcer could be judged by evaluating and documenting the change in the biomechanical response.

This paper introduces a novel handheld instrument, termed pressure ulcer gage, that can detect subtle changes in skin biomechanical properties. The system is inherently precise since there is no sliding surface and because it works in differential mode. It also has a high bandwidth which makes it possible to load the skin statically and dynamically under a wide range of conditions, such as in isotonic or in isometric conditions. By recording and analyzing the response of the skin, it is possible to rigorously monitor the biomechanical properties of the skin. Skin conditions such as stage-I pressure ulcers, deep tissue injury and bruises should be reliably detected. The device is compact, light-weight, robust, and potentially low cost. Therefore, it is suitable for nursing homes.

II. METHODS AND MATERIALS

The instrument comprises a pair of compact, piezoelectric bimorph benders which are arranged to make it possible to achieve large skin strains by pulling the skin from two traction surfaces moving in opposite directions, see Figure 1. The design goals were to develop a portable handheld instrument as per. The specific goals include:

a) Compact and light weight: In most cases, the clinical assessment would be done on immobile patients laying on beds at the hospital or at home. A light weight, hand held device can be manually applied directly to the site under inspection.

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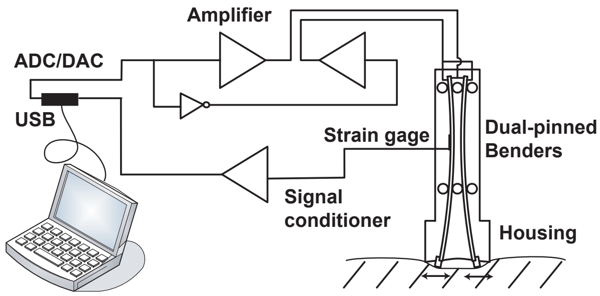

Fig. 1. System diagram.

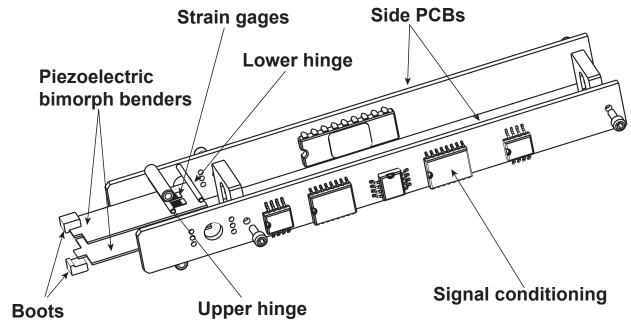

Fig. 2. Exploded view of the pressure ulcer gage

b) Precise and high bandwidth: To detect subtle biomechanical property changes, it is important to precisely and reliably stimulate the skin. Also, a variety of excitation patterns need to be available and the device must have a high bandwidth to load the skin under a wide range of conditions. With various loading conditions, it is possible to test for relaxation, creep, hysteresis, elasticity and so-on.

c) Inexpensive: Since pressure ulcers can be developed quickly under some conditions such as concentrated pressure on the hips, it is important to frequently monitor the skin changes. Therefore, the affordability of the device is very important.

A. Design

Two piezo bimorph benders (Model T220-H4-303Y; Piezo Systems Inc., Cambridge, MA, USA) were mounted to form a pair of tweezer to tangentially stretch the skin as shown in Figure 2. To measure the relative displacement of the benders, two dual grid strain gages (Model EA-30-060PB-350, Vishay Micro-Measurements, Raleigh, NC, USA) were bonded on the both side of the benders, forming a wheatstone bridge. The bridge output was processed by a low distortion instrumentation amplifier (Model INA103; Texas Instruments Inc., Dallas, TX, USA) and filtered by a two-pole active low-pass filter (Model UAF42; Texas Instruments Inc., Dallas, TX, USA) with cut-off frequency at 100Hz. Two high voltage amplifiers (Model OPA445; Texas Instruments Inc., Dallas, TX, USA) were configured in an H-bridge circuit to drive the piezo benders with potential difference between two electrodes varying from -90V to +90V. To eliminate risks of electric shock, two delrin boots were glue on the tip of the piezoelectric benders. The delrin boots were covered with a piece of sandpaper (120 Grate) to increase traction and prevent slipping. To stretch the skin tangentially, a normal force component is necessary. To reduce the variance of the normal force exerted under manual application, a plastic tube was used to guarantee a 1.5 mm indentation of the tweezer tip as shown in Figure 1.

In [8], it was shown that a piezo biomorph bender mounted in a dual-pinned structure is superior than a cantilever structure in terms of stiffness and free deflection. This mounting technique is applied here. To make the device compact, two narrow rectangular printed circuit boards were used as the support for the hinge shafts. The strain gage signal conditioning circuit was laid on these two boards.

B. Constituent equations

Referring to Figure 3, the constituent equation of a dual-pinning installed bimorph piezo bender is given by [8]:

$$\delta = \frac{(l_1+l_2)l_2^2}{2Ewh^3} f^Z + \frac{3d_{31}(l_1+l_2)l_2}{4h^2} V \qquad (1)$$

where δ is the deflection, f^Z is the external force applied at the tip of the bender, E is the piezo material's Young's modulus, h and w are the thickness and width of the layers, d_{31} is the piezoelectric coefficient and V is applied voltage.

Since all governing equations of the bender are linear, at a certain applied voltage, the difference of strain at position l_s is caused by the difference of the external load f^Z. The bending moment at the position l_s is:

$$M = \frac{l_s l_2}{l_1} f^Z \qquad (2)$$

Therefore, the strain difference is found to be:

$$\epsilon_{\text{diff}} = \epsilon_{\text{unload}} - \epsilon_{f^Z} = \frac{M}{EI} = \frac{3}{2}\frac{1}{wh^3E}\frac{l_s l_2}{l_1} f^Z \qquad (3)$$

where $I = 8wh^3/12$ is the moment of inertia of the bender.

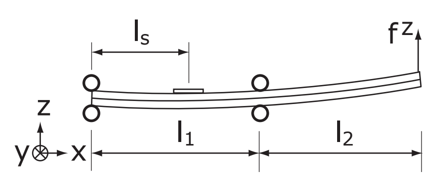

Fig. 3. A dual-pinning installed bimorph piezo bender

C. Calibration

To estimate deflection and external force accurately, Young's modulus E and the piezoelectric coefficient d_{31} of the piezo material must be known. To calibrate these quantities, a dual pinned bender was vertically installed and a tiny mirror was glued at the tip to reflect a laser beam shining on. The deflection of the bender tip was gaged by measuring the displacement of the reflected beam using a lateral position sensing device (PSD, Model DL-10; UDT Sensors, Inc., Hawthorne, CA, USA). The relation between the tip deflection and tip slope of a dual-pinning piezo bimorph bender could be derived as:

$$\theta = \frac{3}{2}\left(\frac{l_1}{2} + l_2\right)\frac{d_{31}V}{h^2} + \frac{1}{4}l_2(2l_1 + 3l_2)\frac{f^Z}{Ewh^3} \qquad (4)$$

The coefficient d_{31} was then calibrated by applying a quasi-static ramping voltage signal while recording the deflection. By applying a known load to the tip of the bender (15 g weight), the Young's modulus was obtained. To minimize errors, the experiments were conducted for l_2 equals to 7, 9, 11, 13, and 15 mm while total length $l_1 + l_2$ is 30 mm. The results agreed to within 5%. The errors caused by actuator hysteresis were compensated by deriving the external force using the strain difference between the loaded and unloaded condition.

III. Experimental results

To verify that the device could reliably detect the disparity of biomechanical properties between skins, several experiments were conducted in which the elasticity and viscoelasticity of forearm skin and palm skin of human subjects were measured and compared. Previous work has shown that biomechanical properties of glabrous skin and hairy skin were quite different [9], [10], [8]. Therefore, a clinically feasible device should be able to consistently distinguish these differences.

A. Subjects

Four healthy subjects, three males and one female, volunteered to participate. The informed consent of the subjects was obtained in accordance with the requirements of the McGill University Policy on the Ethical Conduct of Research Involving Human Subjects.

B. Protocols

1) Quasi-static stretch: A quasi-static ramp voltage varying from -90 V to +90 V was applied to the benders. In this case, the tweezers formed by two benders tangentially stretched the skin from an initial gap of 1 mm. The amplified strain gage signal was sampled at 2 kHz and stored in a personal computer. Before the experiment, the quasi-static ramp voltage was applied when the tweezers were unloaded. Then the strain gage signal was recorded as a reference signal to evaluate skin resistive force according to equation 3. For each subject, four sites on both right forearm and right palm were randomly selected and tested.

2) Sinusoidal loading: A 10 Hz sinusoid signal varying from -90 V to +90 V was applied to the benders. The amplified strain gage signal was sampled at 2 kHz and stored in a personal computer. To eliminate the artifacts caused by the hysteresis property of the piezoelectric material, the sinusoid voltage was applied before the experiment when the tweezers were unloaded and the strain gage output was used as a reference signal in late data analysis. For each subject, four sites on both right forearm and right palm were randomly selected and tested.

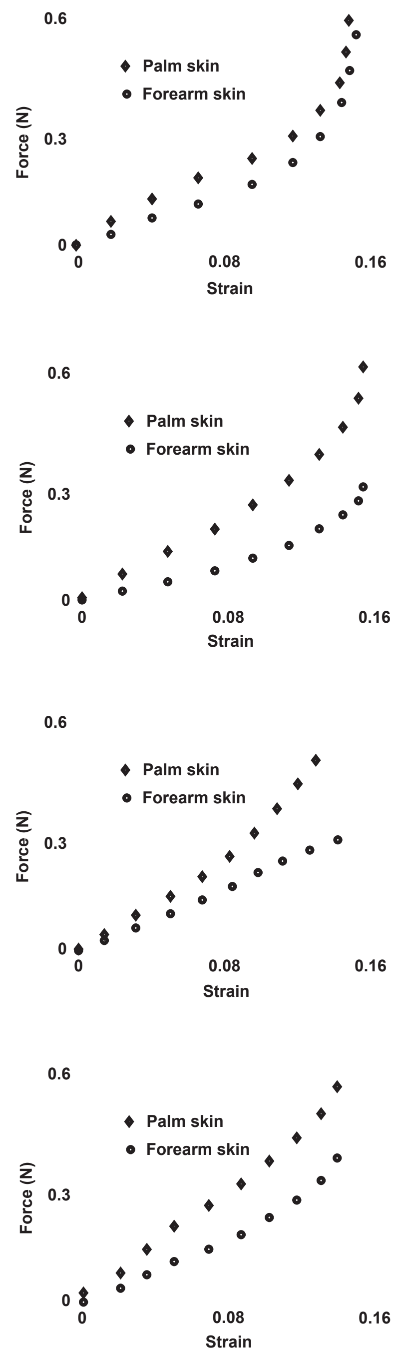

Fig. 4. Force-strain curves of palm skin and forearm skin of four subjects.

C. Results

The results of quasi-static stretch are seen in Figure 4. From the measured force-strain relationship of the hairy skin, i.e. the forearm skin, and glabrous skin, i.e. palm skin, we could see that for some subjects, there were significant differences between the elasticity of the hairy skin and glabrous skin; for others, there were not. However, it is obvious that the hairy skin was consistently softer than the glabrous skin, which agrees with the literature of skin biomechanics [11].

The skin responses to sinusoid signal are illustrated in Figure 5. The phase difference between the input and output results from the viscoelastic properties of the skin. For each subject, the shape of the loop of the forearm skin and the palm skin was different, implying that there were substantial differences in the viscoelastic parameters. In each case, both the slope and the area under the curves could be used as indicators.

IV. Conclusion and Future Work

This paper presents a novel design of a handheld device which could reliably distinguish differences in biomechanical properties of the skin. Preliminary results indicate that this device could be eventually used to detect the onset of ulcers. The device is compact, robust and potentially low cost. The test signals used in the experiments were quasi-static ramps and a 10 Hz sinusoid signal. However, the test signals are not limited to these two cases. For instance, since the constituent equations of the device are known, then it is possible to load the skin isotonically or isometrically by using a close-loop feedback controller. Future work will be directed at finding excitation signals which could maximize the difference between the responses of healthy skin and skin with high pressure ulcer susceptibility.

Since computational capacity needed in the data processing is relatively low, a micro-processor/DSP could be used to replace the laptop computer used here. Moreover, many commercially available general purpose micro-processors/DSP have integrated DAC and ADC. Therefore, the voltage amplifier, data processing and human computer interface could be easily integrated to form a low cost, compact and robust pressure ulcer measuring system.

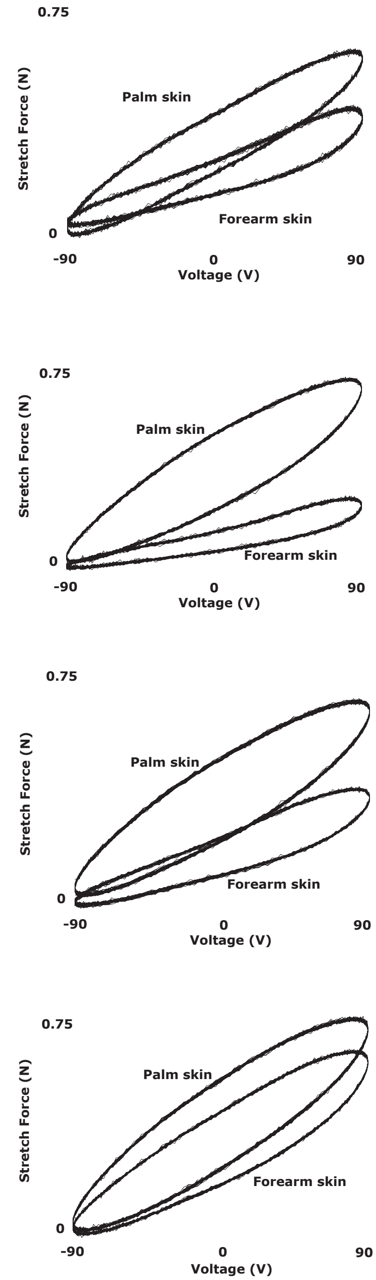

Fig. 5. Viscoelastic behavior of palm skin and forearm skin of four subjects.

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