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Short communication

The effect of biomechanical variables on force sensitive resistor error: Implications for calibration and improved accuracy

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ABSTRACT

Force Sensitive Resistors (FSRs) are commercially available thin film polymer sensors commonly employed in a multitude of biomechanical measurement environments. Reasons for such wide spread usage lie in the versatility, small profile, and low cost of these sensors. Yet FSRs have limitations. It is commonly accepted that temperature, curvature and biological tissue compliance may impact sensor conductance and resulting force readings. The effect of these variables and degree to which they interact has yet to be comprehensively investigated and quantified. This work systematically assesses varying levels of temperature, sensor curvature and surface compliance using a full factorial design-of-experiments approach. Three models of Interlink FSRs were evaluated. Calibration equations under 12 unique combinations of temperature, curvature and compliance were determined for each sensor. Root mean squared error, mean absolute error, and maximum error were quantified as measures of the impact these thermo/mechanical factors have on sensor performance. It was found that all three variables have the potential to affect FSR calibration curves. The FSR model and corresponding sensor geometry are sensitive to these three mechanical factors at varying levels. Experimental results suggest that reducing sensor error requires calibration of each sensor in an environment as close to its intended use as possible and if multiple FSRs are used in a system, they must be calibrated independently.

1. Introduction

Quantifying biomechanical forces between medical devices and human soft tissue has important implications for comfort, reducing tissue injury and improving device design (Dabling et al., 2012; Lebosse et al., 2011; Mak et al., 2010). Typical measurement of these interactions requires a sensor positioned at the interface between the tissue and medical device. Many biomechanical sensors are described in literature including those based on capacitance, fluid pressure, or optics (Dabling et al., 2012), with one of the more prevalent sensors being Force Sensitive Resistors (FSRs). FSRs are constructed of thin polymer films and change resistance with the application of pressure. With sensor thicknesses as little as 0.2 mm (Dabling et al., 2012), FSRs can be positioned between two contacting surfaces with little mechanical impact on the substrates. FSRs require minimal signal conditioning and are easily integrated with hobbyist micro-controllers through advanced data acquisition systems. FSRs are inexpensive compared to similar technologies (Dabling et al., 2012; Lebosse et al., 2011) making them an attractive option for research and clinical applications.

FSRs have been employed in numerous biomechanical applications from prosthetic control and pressure measurements (Hebert et al., 2014; Junaid et al., 2014; Silver-Thorn et al., 1996) through gait studies (Moon et al., 2011; Rueterbories et al., 2010) and telerobotics (Yun et al., 1997) among many other biomechanical applications quantifying interface mechanics (Cascioli et al., 2011; Di Fazio et al., 2011).

However, FSRs have limitations; sensor drift and hysteresis have been shown to impact repeatability and accuracy (Dabling et al., 2012, Herbert-Copley et al., 2013). Additionally, changes in accuracy and increases in drift error when curvature is applied to the sensors have been shown in prosthetic applications (Polliack et al., 2000a), but can be minimized by calibration in the same curved configuration (Buis and Convery, 1997).

FSR manufacturers often recommend calibration and operating conditions to include flat, rigid surfaces at room temperature (Interlink Electronics, 2015; TekScan, 2015). Yet the human body

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hosts unavoidable curvatures, soft tissue compliances, and temperature differentials. The error imparted by these variables has yet to be comprehensively investigated, preventing researchers and clinicians from understanding the implications of their biological testing environment on sensor accuracy.

2. Objectives

This work investigates the effects of common biomechanical variables on FSR error with the intent of examining calibration practises and providing recommendations to improve accuracy in a clinical-research environment.

3. Methods

3.1. Variable testing

3.1.1. Experimental variables

A full factorial design-of-experiments approach was used (Montgomery, 2012). Twelve unique combinations of temperature, curvature and compliance were introduced to each FSR in a semi-randomized order (Table 1). Temperature was evaluated at room (21 °C) and body (37 °C) temperature; curvature at the diameter of a 95th percentile male thigh (215 mm), diameter of a 5th percentile female wrist (44 mm) (NASA, 2008) and a flat surface; and material compliance of a human soft tissue analog (SynDaver Labs, Tampa, USA) and a rigid surface.

3.1.2. Setup and procedure

Interlink FSRs were selected for testing due to their widespread usage (Hebert et al., 2014; Jang et al., 2010; More and Lka, 2014; Rogers et al., 2010; Wang et al., 2010; Yun et al., 1997). Two small round (5 mm diameter), two medium round (13 mm diameter) and two 38 mm square (Models 400, 402 and 406 respectively, Interlink Electronics, Camarillo, USA) FSRs were tested. Once calibrated, manufacturer specifications state force accuracy in a range of ±6% to ±50% (Interlink Electronics, 2015). FSRs were wired to a data acquisition system (PCI 6259, National Instruments, Austin, TX, USA) connected to a 10 kΩ resistor in a voltage divider configuration (Interlink Electronics, 2015). FSRs were placed in-line with a load cell calibrated to an accuracy of ±0.02 N (LCM703, Omegadyne, Sunbury, USA) affixed to a micromanipulator (MM-3, Narishige Group, Tokyo JA) (Fig. 1). Custom PLA-thermoplastic pushing heads were 3D printed to match the sensing surface dimensions of each FSR and introduce curvature as required (Fig. 1). During testing, FSRs were pressed between the pushing head and the test surface. The testing assembly was located inside an incubator (Air-Shields C100, Soma Technology Inc., Bloomfield USA) allowing for precise temperature control.

Although the FSRs selected have a working range between 0 and 20 N of force (Interlink Electronics, 2015), a testing range of 0–10 N was used (Hollinger and Wanderley, 2006). The upper bound was limited to 10 N, as further force can cause discomfort if applied to human soft tissue over a small surface area (Antfolk et al., 2010; Armiger et al., 2013). Data collection was conducted according to ANSI/ISA 51.1 Standards (ANSI, 1995). Accordingly, FSRs were preconditioned and data logging was initiated mid-way through the force range. The FSRs were loaded to the maximum and minimum values three times at a consistent loading rate (Interlink Electronics, 2015) of 30 s/cycle. This loading rate was chosen to reflect a low frequency or static application and to avoid any time dependent dynamic effects (Interlink Electronics, 2015; Lebosse et al., 2011). FSR voltage and load cell forces were sampled at 100 Hz, low-pass filtered at 20 Hz and 10 Hz respectively, and logged at 10 Hz.

Table 1
Combinations of biomechanical variables tested.

Combination	Temperature (°C)	Curvature (Diameter mm)	Compliance
1	21	215 mm	Rigid
2	21	44 mm	Rigid
3	21	Flat	Rigid
4	21	215 mm	Soft
5	21	44 mm	Soft
6	21	Flat	Soft
7	37	215 mm	Rigid
8	37	44 mm	Rigid
9	37	Flat	Rigid
10	37	215 mm	Soft
11	37	44 mm	Soft
12	37	Flat	Soft

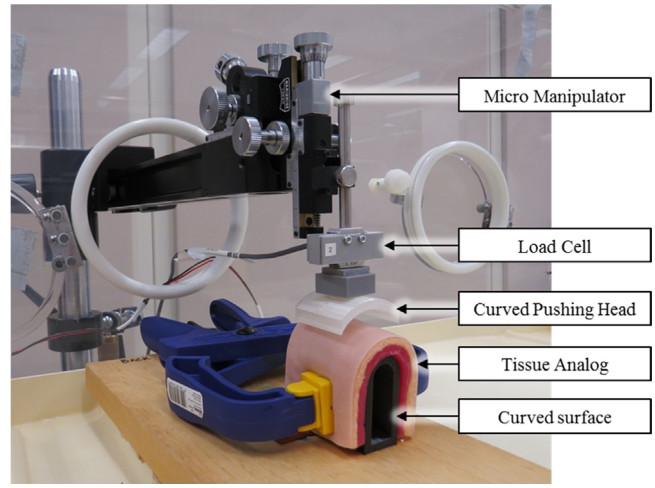

Fig. 1. Experimental setup. Experimental setup for testing of the 12 combinations of variables. Setup shown in the body temperature, 44 mm diameter, and soft compliance configuration.

3.1.3. Data treatment

For each FSR, calibration equations mapping FSR voltage to applied load (load cell reading) were determined through fitting an inverse logarithmic equation (Eq. (1)) as recommended (Interlink Electronics, 2015).

$$F = ae^{bV} + c \tag{1}$$

where F represents the force predicted from the calibration equation, V measured voltage from the FSR and a, b, and c are constants to be solved for each sensor and combination of variables.

Twelve equations per FSR were determined corresponding to the twelve combinations of temperature, curvature and compliance introduced. The fitted-root-mean-squared-error (RMSE-F), mean absolute error (MAE), and maximum error were calculated and recorded for each combination (Supplementary Table 1).

Data for each combination of biomechanical conditions was evaluated under three calibration strategies: ***self***-fit calibration, each sensor calibrated 12 times, once for each combination of variables; ***baseline***-fit, often recommended by manufacturers (Tekscan, 2015), each sensor is calibrated once under optimal conditions (flat, rigid, and room temperature); and ***cross***-fit, one baseline calibration equation applied to all sensors of the same model. Mean differences in RMSE-F, MAE, and maximum error, corresponding to calibration fit strategy, were determined and statistically compared using paired t-tests, with $p < 0.05$ indicating significance.

The Baseline calibration equation for each sensor was defined as the flat, rigid, room temperature condition. At each of the remaining 11 combinations, calibration equations were compared to the baseline using root means squared error (RMSE-C). This procedure was performed for each sensor independently, yielding twelve RMSE-C values for each of the six sensors. A graphical example is illustrated in Fig. 2.

Three analyses of variance (ANOVAs) were performed, one for each sensor model. Since two sensors of each model were used, this data was treated as replicate measures in the statistical analysis. Temperature, curvature and compliance were held as input variables with RMSE-C evaluated as the output measure, and blocking performed by sensor number. Initially, all main effects, 2-way and 3-way interactions were evaluated with $p < 0.05$ indicating significance. Non-significant variables were then removed from the model. Significant main effects, significant 2-way interactions, and the main effects corresponding to any significant interactions were reported.

3.2. Participant testing

Two healthy participants were recruited. Ethics approval was obtained through our institute's review board and participants gave written informed consent.

Participant testing closely paralleled the variable testing procedure described previously and was intended to simulate the implications of FSR usage in a biomechanical system. Participants' arms were secured using an adjustable arm rest. Each FSR was adhered directly to the participants' skin and given minimally 15 min to reach a stable temperature (approximately 32.5–34 °C). Using the previously described preconditioning and loading procedures, the micromanipulator, load cell, and FSR pushing heads were then pressed tangentially onto each participant's forearm directly over the sensor (Fig. 3). FSR and load cell data was captured over a 0–10 N range for each FSR (small round FSRs limited to 0–8 N to reduce discomfort).

From this data, calibration equations were derived for each FSR using the same inverse logarithmic equation described previously. Comparing participant data against each sensor's previously derived baseline equation; differences in mean RSME-C, MAE and Maximum error were evaluated using a paired t-test, with $p < 0.05$ indicating