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Multi-Axis Foot Reaction Force/Torque Sensor for Biomedical Applications

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***Abstract*—To support an Oak Ridge National Laboratory programs exoskeleton project, a unique multi-axis foot force/torque sensor was constructed and tested that has biomedical application such as clinical gait analysis. The challenging aspect of this multi-axis force sensor is that it had to conform to the bending of the human foot, withstand high impact loads, have high force sensitivity, feel comfortable to the human wearer, be integrated into a military style boot, measure the forces on the human foot when either the ball or the heel of the foot is in contact with the ground, respond to both positive and negative loading and have a low overall height and weight. This paper describes the design and testing of this unique sensor.**

I. Background

SENSING forces and motion of humans is required in a number of areas ranging from the construction of legged robotic systems and exoskeletal systems, to the study of human body movements such as in gait analysis. While the first two areas obviously require portable force sensing, researchers in the area of human body movement and biomechanics commonly utilize stationary force plates to capture ground reaction forces on each foot. A force plate poses restriction on the natural gait pattern and range of activities that can be tested (e.g., jumping and walking outdoors). To overcome these limitations, a number of researchers have been constructing portable foot sensors. A number of approaches are typically taken in the design of a foot sensor. One approach is to use a rigid plate with strain gauges attached (see Ref. 1). These sensors are then attached to the shoe for the subject to wear. While good sensing accuracy can be achieved with this design, the rigid attachment interferes with natural foot motion and feel. To broaden the gait analysis and comfort, conventional running shoes are modified (see Refs. 2-5) to include force sensitive resistors, capacitor sensors, bend sensors and pressure sensitive materials to detect heel and toe strike. This approach, while inexpensive, typically cannot capture all the forces and moments exerted on the foot. Forces are typically only compressive and only measured in a few axes with low precision. Other types of sensors can be added to a foot sensor such as velocity-based gyros, accelerometers and inclinometers (see Refs. 6 and 7) in an attempt to augment or replace video capture data for gait analysis. But the fundamental problem of collecting accurate force/torque measurements isn't aided by these additional sensors. Finally, dual multi-axis force/torque sensors placed at the heel and toe areas are sometimes used (Ref. 8). While such systems add measurement axes they are relatively tall and have redundancy of measurement when both heel and toe sensors are in contact with the ground. The relationship of the frame of reference between the two force/torque sensors is also difficult to determine due to the flexing of the sole of the shoe.

As part of an Oak Ridge National Laboratory program to develop a powered human exoskeleton, a new type of foot reaction force/torque sensor was developed that overcomes shortcomings of previous designs. In this new design, load cells are interposed between closely spaced flexible plates in such a way as to accurately measure the full 6 axes of force and torque at the foot while retaining a comfortable feel. This flexible sensor attaches to sole of a modified boot and adds only 2.6 cm of height and 839 grams of mass. It is capable of over 5000 N of total vertical load. While specifically targeting exoskeleton needs, the basic design of the foot sensor is applicable to biomedical measurement.

II. Description of Foot Sensor

A. Design Parameters

The design targets met for this sensor are listed in Table 1. As noted from the table, vertical, lateral and moment loads can be measured. A low overall sensor height and mass was targeted to maintain a more natural feel to the user.

B. Mechanical Description

The foot sensor utilizes two flexible plates that are constrained to maintain similar curvature. The top plate is in contact with the human boot sole while the bottom plate contacts the ground (see Figs. 1 and 2). The plates are thin and flexible enough so that normal foot flexing is permitted. The upper plate is attached to the human boot through bindings so that all forces and moments generated by the foot are transmitted to the upper flex plate. The boot binding, which is tight at the ball of the

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foot, permits a small amount of fore/aft displacement between the boot sole and flex plate at both the heel and toe. This displacement allows normal flexing of the foot without stiffening the boot. The upper flex plate is connected to the lower (ground contact) flex plate through a number of vertical and horizontal force sensing elements (see Fig. 3). These connections are designed to transmit the forces to the sensing elements in their sensitive axis while not imposing loads in unmeasured directions. The design of these connections is very important to the overall accuracy of the foot sensor system. A key advantage of this design is that the number of required force sensing elements is reduced since the flex plate carries all the transverse loads (lateral and fore/aft). In this design, some translational displacements occur between the upper and lower flex plates during flexing. These relative motions occur primarily in the X direction and are accommodated by the connections between the upper flex plate and the force sensors that are mounted to the lower flex plate. Z-direction loads imposed on the flexible foot input plates are transmitted to the load sensors through flexible linkages that allow significant angular and X translational motion (with little resistive force) while maintaining stiffness in the load sensitive Z-direction. Lateral constraint elements are used to minimize lateral forces on the vertical load cells which helps assure accuracy of load measurement. The lateral constraint elements behave like ball bearings so that nearly friction-free flexing and translation of the flexible input plate is permitted without lateral displacement.

TABLE I
FOOT SENSOR DESIGN ACCOMPLISHMENTS
(SEE FIGURE 1 FOR REFERENCE FRAME)

1. Maximum Total Vertical (Z) Force	5,340N (or 1,200 lbf)
2. Maximum Fore/Aft (X) Force	890 N (or 200 lbf)
3. Maximum Heel Lateral (Y) Force	445 N (or 100 lbf)
4. Maximum Ball Lateral (Y) Force	445 N (or 100 lbf)
5. Maximum Moment (My)	373 N-m (or 3,300 in-lbf)
6. Maximum Heel Moment (Mx)	59 N-m (or 525 in-lbf)
7. Maximum Ball Moment (Mx)	80 N-m (or 690 in-lbf)
8. Maximum Moment (Mz)	62 N-m (or 550 in-lbf)
9. Maximum Toe Force	445 N (or 100 lbf)
7. Total Height	2.6 cm [or 1.03 in (=0.73” sensor height + 0.3” bottom tread thickness)]
8. Sensor Mass	839 grams (or 1.85 lbs)
9. Force Resolution	1.1 N (or 0.25lbf per load cell)
10. Minimum Radius of Curvature	30 cm (or 12 in)

Figures 1 and 2 show the sensor attached to a military boot in the flat configuration (it is 2.6 cm thick including the 0.8 cm rubber padding on the bottom).

Fig. 1. Foot force-torque sensor attached to boot.

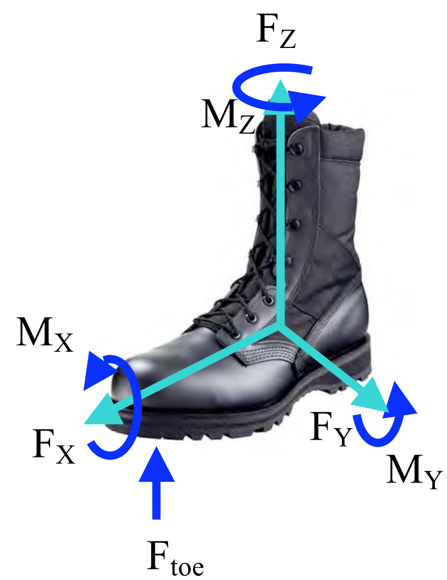

Fig. 2. Foot force-torque sensor coordinate system.

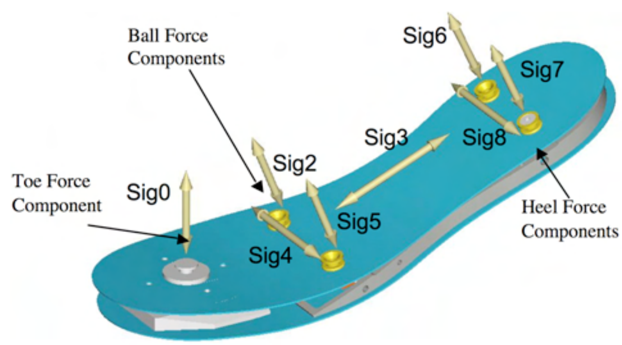

Fig. 3. Sensor force measurements.

The individual components of force are measured with strain gage load sensing elements which are built into frames between the upper and lower flex plates. Both bending beam and shear type load sensing elements are employed and they are each individually calibrated with test masses before final assembly. To ensure proper alignment, test fixtures were developed for calibration of each load cell. At least 10 test masses, ranging by 2 orders of magnitude from the peak load, ensured linearity of each load cell. High Z-direction loading of the sensor can be concentrated at the heel or ball so the load cells at positions 2, 5, 6 and 7 were designed for up to 1300 N each. This results in the relatively high total Z-direction load capacity of 5340 N. Figure 4 shows the internal arrangement of the load cells. Full Wheatstone bridges are employed using temperature compensated strain gages. The degree of bending in the upper flex plate is also measured with strain gages. This allows the off-axis components of force created by the angled contact to be calculated and factored into the output results resulting in a more accurate computation of forces and moments. The mapping of the strain gauge signals (Sig0 to Sig8) can be shown to be

$$\begin{bmatrix} F_x \\ F_y \\ F_z \\ M_x \\ M_y \\ M_z \end{bmatrix} = \begin{bmatrix} Sig3 - Sig0\sin\theta \\ Sig4 + Sig8 \\ Sig2 + Sig5 + Sig6 + Sig7 + Sig0\cos\theta \\ .875(Sig7 - Sig6) + 4(Sig4 + Sig8) + 1.125\,Sig5 - 1.075\,Sig2 \\ -5.5(Sig2 + Sig5) - 8Sig0\cos\theta + A(Sig0\sin\theta - Sig3) \\ .62\,Sig8 - .42\,Sig3 + 5.5\,Sig4 \end{bmatrix}$$

where (0,0,0) is at the center of ankle rotation (10 cm above the foot sensor) and theta is the angle of the toe force (theta=0 when the sensor is flat). A final series of calibration tests were conducted on the total sensor verifying linearity and independence (lack of cross-axis coupling).

C. *Electronic Description*

Some details of the strain gauge sensing elements are shown in the photos (Fig. 4). The strain gage wiring is routed to a low-level data acquisition board mounted in the load sensing frame.

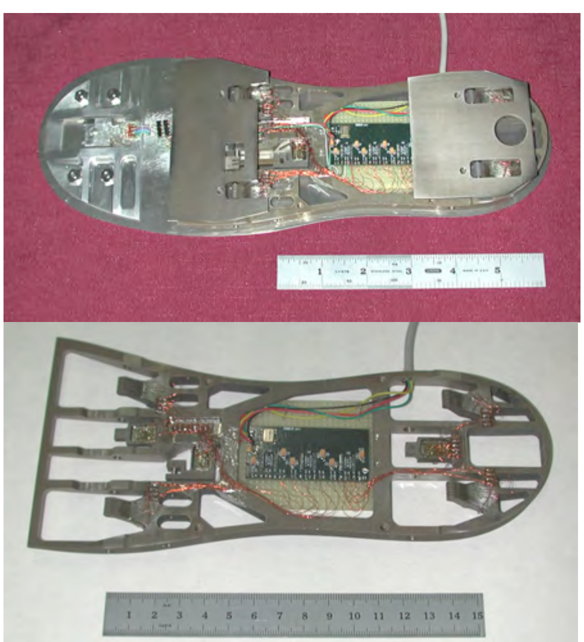

Fig. 4. Sensor load cell details.

The low-level data acquisition system, implemented on a custom circuit board embedded within the foot sensor, is shown in Fig. 5. It is based on the AD7731 chip, a low noise, high throughput 24-bit delta-sigma ADC. Each chip features three buffered differential programmable gain inputs that can be interfaced directly to the strain gages. Since the ADC chip and the strain gages share the same 5 volt power supply, small variations in supply voltage affect both the strain gauges and the ADC thus minimizing supply related errors. Three of the chips are required to interface to a total of 9 strain gauges in the foot sensor. A PIC16F873A microprocessor controls the three AD7731 chips and reads the digitized data utilizing a high speed SPI serial link. Temperature is sensed using a DS18B20 chip and read by the microprocessor through a simple 1-wire interface. This signal is used to correct for temperature drift associated with the strain gauges. The microprocessor then assembles the data into a single packet to be uploaded to an external computer (currently a PDA) at an update rate of 230 Hz through a standard RS-232 serial link for display and storage. The microprocessor is configured to boot from internal flash memory so the only connections required are for power and the serial link to an external computer. The current configuration utilizes an external 9 volt battery, regulated to 5 volts for power. The AD7731 chips can be configured to trade off noise and resolution against throughput by varying the internal filtering. For the current application, the filtering is configured such that a throughput of up to 1,200 Hz is available. This corresponds to a -3 dB frequency of 314 Hz and a peak-to-peak resolution of 13.5 bits. The update rate to the PDA was limited to 230 Hz because of the limited ability of the PDA to process the incoming data at faster speeds.

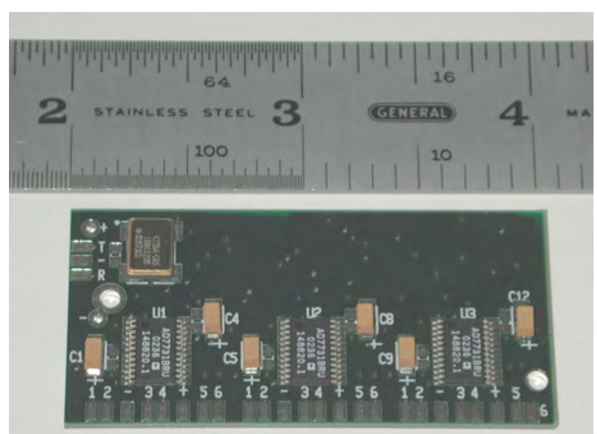

Fig. 5. Load cell signal conditioning board.

III. FOOT SENSOR TESTING

After calibrating the individual load cells (described in the previous section), data was collected from the sensor during standing, walking (slow and fast) and jogging. Figures 6 through 10 show typical data collected in these activities. The foot sensor data compares favorably with published force data that has been taken from commercial, stationary force plates (see Ref. 9). Static loading of the sensor by a test subject of known weight added an additional validation of the accuracy of the systems sensitivity and accuracy. Only one foot has the sensor with the other foot utilizing a dummy insole.

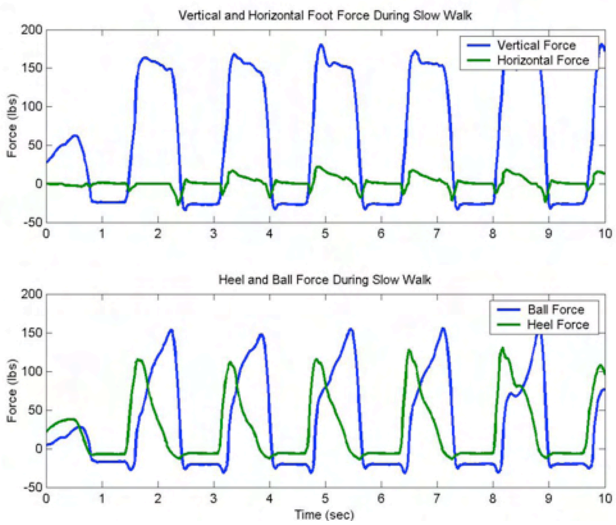

Fig. 6. Slow walk force data (magnitudes only).

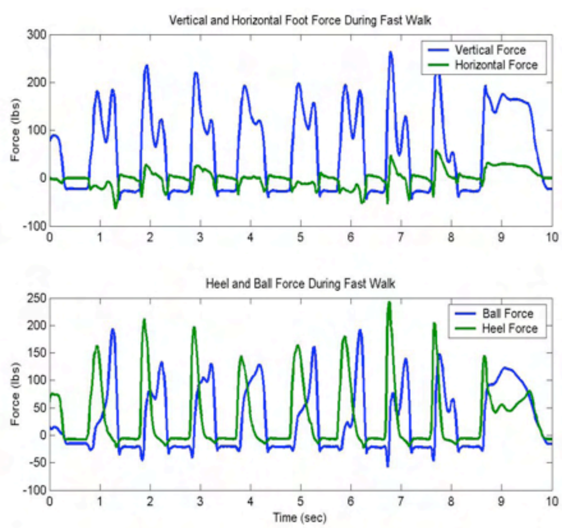

Fig. 7. Fast walk force data (magnitudes only).

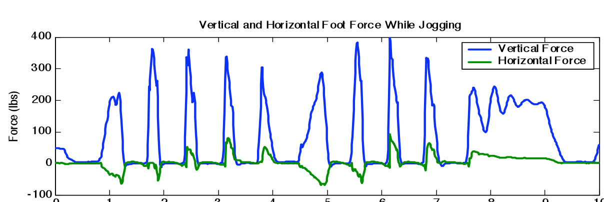

Fig. 8. Jogging force data (magnitudes only).

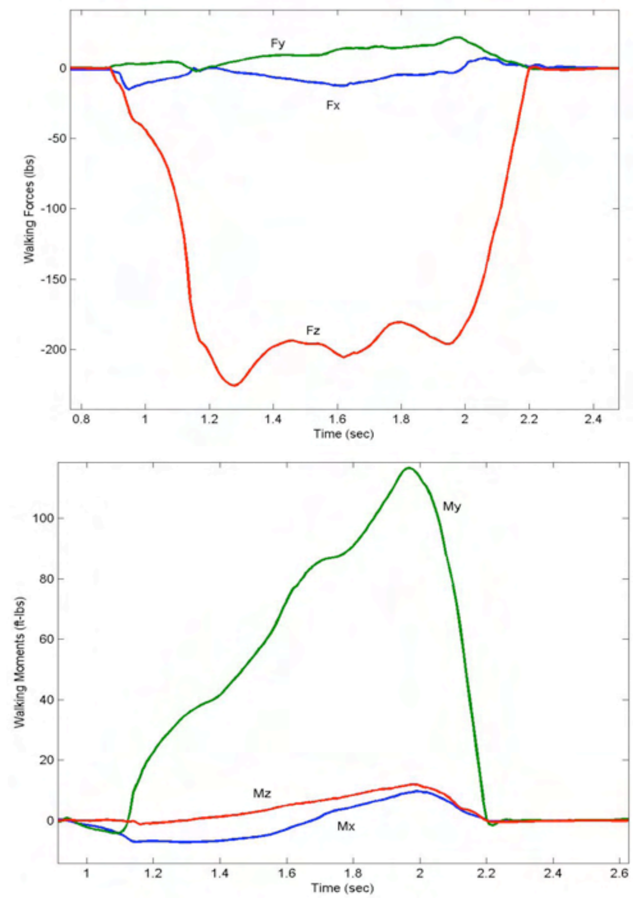

Fig. 9. Individual forces and moments measured during walking (slow).

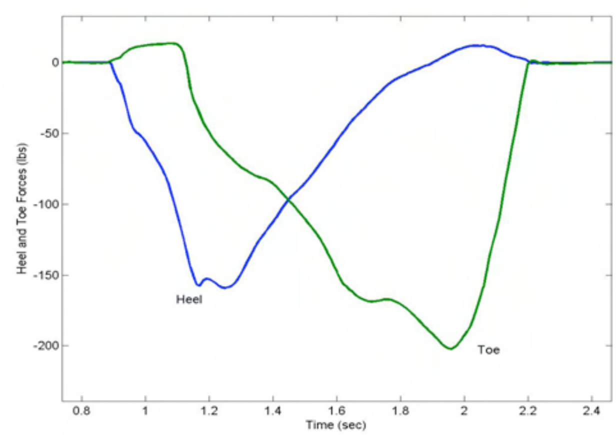

Fig. 10. Toe and heel forces during walking.

Because of the flexing of the insole and low mass (in comparison to the boot mass), comments from users of this device has been very positive. The data collection is done by means of a PDA, which can be attached to the user's belt or carried in a pocket. Externally, only a single 9V battery in a small case including a voltage regulator, power switch, power indicator and serial interface is used. Its placement is again easily accommodated in the user's pocket or strapped to his/her belt.

IV. Summary

A flexible foot reaction force sensor has been developed utilizing a unique arrangement of strain gage load cells. The wearable sensor is lightweight with a low profile and is designed to attach directly with a typical boot (with minor modifications). The sensor is rated for high normal and moment loads that may be encountered in jumping and running. The sensor provides data similar to stationary force plates without constraining the subject to step in a pre-defined spot and provides continuous real-time sensing profiles. By using highly accurate strain gage load cells that are widely separated and coupled to minimize off-axis loading, good overall accuracy and sensitivity is achieved. The design provides space inside the sensor for integration of signal conditioning electronics to provide serial data output. The sensor is readily sealable to prevent moisture and environmental contaminants from entering the device. While the sensor described in this paper was designed for exoskeleton requirements, the basic design can be scaled either up or down.

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