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Instrumented Insole vs. Force Plate: A Comparison of Center of Plantar Pressure

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***Abstract*—Instrumented insoles allow analysis of gait outside of the confines of a motion analysis lab and capture motion data on every step. This study assesses the concurrent validity of center of plantar pressure (COPP) measurements during walking, and shows that our custom instrumented insoles compare favorably to an Advanced Mechanical Technology Inc. (AMTI) force plate in a clinical motion laboratory, particularly when the large difference in price is considered (an insole is nearly two orders of magnitude less expensive than a force plate). Deploying inexpensive insoles such as ours for ubiquitous health monitoring allows measurement of gait in more typical environments. This affords the opportunity to evaluate the gait of older adults in the home environment, and a future opportunity of providing real-time feedback corresponding to changes in gait.**

I. Introduction

FALLINGis a major cause of death and health problems for the elderly population. In 2000, the population over age 65 experienced 10,300 falls resulting in death and 2.6 millions falls resulting in injury; the associated direct health-care costs were $0.2 billion USD and $19 billion USD respectively [1]. Falls have been an increasing cause of death since 2000, and the most recent data from 2007 shows the population over age 65 experiencing 18,334 falls resulting in death [2]. As the elderly population increases, the healthcare costs associated with falls will increase as well. Intervention is necessary if these increasing costs are to be reduced.

To assess instability and fall risk, clinicians typically test patients by having them perform simple motor tasks (e.g. Berg Balance Scale and Functional Reach) or rise out of a chair and walk a short distance (e.g. Timed Up and Go). Although clinical evaluation is practical, it provides only a brief snapshot of a patient's abilities, disregarding the complex gait activity found in the home environment. Indeed, Functional Reach has been shown not to reliably measure dynamic balance [3], and the Berg Balance Scale and Timed Up and Go tests also failed to predict fall risk [4]. To properly evaluate dynamic gait stability, measurements made in the home environment are needed.

One aspect of gait that holds promise as an objective measure of a patient's gait stability is center of plantar pressure (COPP). One research group has shown that COPP characteristics add predictive power to a constructed index for evaluating dynamic gait stability [5], [6]. Likewise, other groups have taken an interest in exploring the usefulness of COPP for indicating aberrant gait [7], [8]. Strictly speaking, plantar pressure distribution, not COPP, has been used in several studies to indicate gait instability by providing sufficient information for identifying differences between persons with and without PD [9, 10] and between elderly fallers and non-fallers [11]. Similarly, other studies have shown that plantar pressure distribution is a reliable indicator of instability for persons with ankle problems [12] and hemiparetic patients [13].

At present, there are a few devices that are capable of measuring plantar pressure, and more specifically COPP. The current gold standard is a force plate, but this has not been and most likely cannot be acclimated for home use. Tekscan makes various pressure sensitive insole systems under the name of F-scan, but all of these are prohibitively expensive for widespread consumer or clinic use, costing upwards of $10,000USD per pair. In addition, both Tekscan and the Parotec system developed by London Orthotic Consultancy, require tethered data lines to a waist unit making it highly obtrusive [14]. novel is another company that also makes a pressure sensitive insole device, costing upwards of $20,000USD per pair, which is very similar to the F-scan line, with the same limitations [15]. Wertsch et al. created insoles capable of detecting pressure under specific areas of the foot, and although this system was untethered, the sampling rate allowed for a data collection period of only five seconds per minute [16]. An intriguing design using air pressure sensors was incorporated into the Berkeley SmartShoe, but this system requires drastic modifications to a user's shoes as well as a tether for capturing data [17].

Observing these limitations, we have developed an inexpensive, unobtrusive, untethered insole-based system that can collect plantar pressure data continuously for over ten hours. We have previously discussed the concept of "just enough measurement," suggesting that a large number of inexpensive measurements (e.g. as obtained by an insole in the home environment) can provide meaningful medical results comparable to or exceeding single high-quality measurements (e.g. as obtained in a motion laboratory) [18].

This paper assesses the concurrent validity of the COPP calculated from this system with the COPP calculated from a

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force plate. Section II discusses the methods, followed by results and discussion in Sections III and IV, and concluding remarks are presented in Section V.

II. Methods

A. Hardware

A custom instrumented insole was constructed by embedding force sensitive resistors (FSRs) from Interlink Electronics and custom printed circuit boards (PCBs) in a layer of polydimethylsiloxane (PDMS) (see Fig. 1). The PCB in the toe piece was designed for placing multiple FSRs under the first and fifth metatarsal head regions to facilitate use for different foot sizes in the range US Men's 5.5-11.

Wires carrying FSR signals from the PCBs are connected to a TI ez430-RF2500 development board housed in an ankle-mounted box. Powered by one AA battery, the insole circuitry reads and wirelessly sends data at over 100Hz to a nearby computer where data is stored and processed [19].

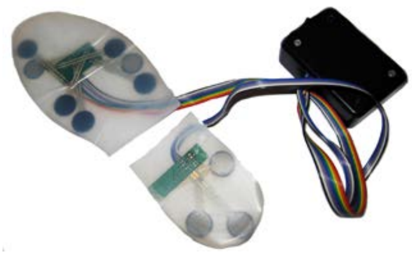

Fig. 1. Custom-built instrumented insole.

B. Experimental Procedure

Our insole was taped down to an AMTI force plate [20], and one FSR was tapped three times. One healthy 28-year-old test subject stepped on the insole 50 times (lining up his right foot on the insole as best as he could) while making sure the left foot did not touch the force plate. After the steps, the FSR tapped prior to the 50 steps was tapped three times again. Force plate data (COPP position and force) and insole data (forces on FSRs) were recorded.

Tapping one FSR before and after data collection allows the force plate data and insole data to be lined up in time; in this way, each insole reading can be paired with the corresponding reading from the force plate. Coordinates of the FSRs were determined by pushing on each FSR separately for two seconds and using the resulting coordinates reported by the force plate.

C. Calibration

Data recorded from the FSRs are in the form of 10-bit analog-to-digital (A2D) readings. Behavior among FSRs varies; applying equal forces to different FSRs does not entail equal A2D readings. In order to calibrate each FSR, a 50lb loadcell from Loadstar Sensors [21] connected to 0.5" diameter cylindrical attachment was fastened to a vise, as shown in Fig. 2. The attachment was centered over each FSR and the vise was closed gradually, applying force to the FSR through the loadcell, while a custom MATLAB script recorded outputs from the loadcell and FSR. Calibration data was recorded for forces ranging from 0lb to 30lb.

Fig. 2. Loadcell mounted in a vise for calibration of FSRs

D. Data Analysis

Coordinates of the COPP for each step were calculated by first finding heel-strike (HS) and toe-off (TO) points which indicate when a step begins and ends (i.e. stance phase of gait), respectively. HS points were determined as the points the force curve increased from the baseline unloaded level. TO points were the subsequent points where the force curve decreased back to the baseline level. Stance time for each step was calculated by subtracting HS from TO. Each COPP coordinate was calculated as the weighted mean of the FSR positions, using the forces on each FSR as the weights. These coordinates were compared with coordinates reported by the force plate using Pearson's Correlation Coefficient. In addition, the RMS error was calculated for X and Y positions separately.

Two spatiotemporal variables are of interest for use in gait analysis: path distance and roll speed. Path distance is calculated as the cumulative sum of the differences between sequential COPP coordinates during stance. Roll speed is calculated as the mean speed at which the COPP travels during stance.

III. Results

A. Calibration

The forces recorded from the loadcell plotted against the A2D output recorded from each FSR during the calibration process are shown in Fig. 3. The least-squares-fit curves are indicated by black lines.

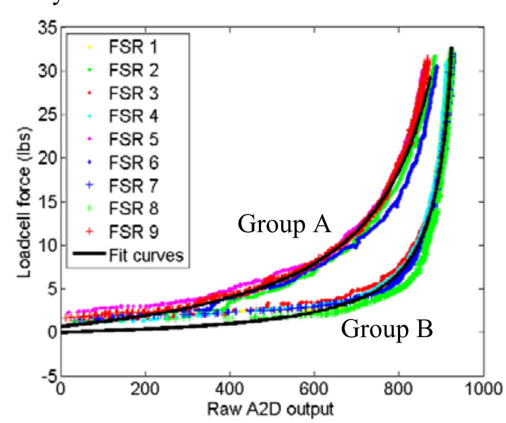

Fig. 3. Calibration data for FSRs. The behavior indicates two groups (A includes FSRs 2, 5, 6, and 9, and B includes FSRs 1, 3, 4, 7, and 8).

B. Comparison of COPP

The COPP for three of the 50 steps as reported by the force plate data and as calculated from the insole data are shown in Fig. 4.

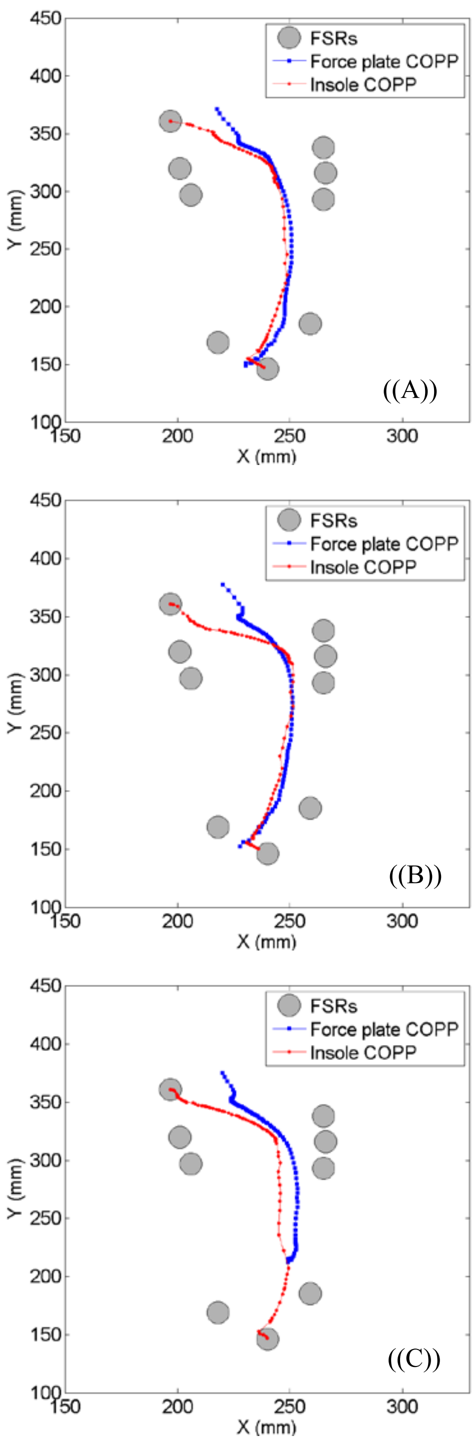

Fig. 4. COPP as measured by the instrumented insole and the AMTI force plate for three walks: a) the walk with the minimum RMS error in x and y, b) a representative walk with RMS error close to the mean in x and y, and c) the walk with the maximum RMS error in x and y.

Table I lists the Pearson Correlation Coefficients and the RMS error comparing the insole-derived COPP coordinates with those reported by the force plate.

Table I
Pearson Correlation Coefficients and RMS Error

	Pearson Correlation		RMS (mm)	
	X	Y	X	Y
Min	0.68	0.97	7	13
Max	0.97	0.99	14	24
Mean	0.87 ± 0.07	0.99 ± 0.003	11 ± 1	17 ± 2

C. Comparison of Spatiotemporal Variables

Table II shows the results for path distance and roll speed as calculated from the force plate and the insole, along with the percent change using the force plate value in the denominator.

Table II
Differences of Spatiotemporal Variables

	PathDistance (%)	RollSpeed (%)
Min	11	-4
Max	40	33
Mean	28 ± 6	19 ± 7

Over all 50 steps, the stance time calculated by the insole differed from the force plate by an average of 83 ± 15ms, or 9 ± 2%.

IV. Discussion

A. Calibration

The variation of FSR behavior is readily apparent. The data indicates that these FSRs can be divided into two groups: four which gently increase in force over an A2D domain of ~100 to ~875, and five which seem to have an effective domain of ~650 to ~925. Instead of calibrating each FSR individually, the data in each group was used to find two calibration equations were generated using a least-squares-fit in MATLAB. Fig. 3 shows the grouped curves with the best-fit curves overlaid in black.

The major contributor to the presence of two groups is likely the characteristics of the sensors due to manufacturing techniques; the response to applied forces will vary between FSRs even without embedding them in PDMS. The insole layout may also contribute; PDMS will expand outward when squeezed, and the presence of the PCBs may impede some of this expansion when squeezing areas close to the PCB. Indeed, the FSRs in one group are those whose active areas are within a distance of 8mm from a PCB whereas the other group falls within distances >15mm.

Calipers were used to measure the thickness in millimeters of the PDMS over each FSR. One group's thicknesses were 5.0, 5.1, 5.4, and 5.4 whereas the other group's were 4.9, 4.9, 5.0, 5.2, and 5.3, suggesting that PDMS thickness was not a factor in creating different behavior.

B. Comparison of COPP

For each step, the COPP coordinates generated by the force plate and those generated by the insole were compared by calculating the Pearson correlation coefficient and the

RMS error. Good correlation (ranging from 0.68 to 0.97) was found for the X-coordinates with 40 of the 50 steps having a correlation of 0.80 or greater. Very good correlation (ranging from 0.97 to 0.99) was found for the Y-coordinates. With all the steps analyzed collectively, the correlation coefficients of the X- and Y-coordinates were found to be 0.87 and 0.99 respectively.

Whereas a force plate is used to analyze individual steps, and thus needs to possess little error in calculating COPP coordinates, the insole is intended for analyzing trends over time periods of many hours or days. As the correlation coefficients show, the insole correlates well with a force plate over 50 steps, which is what would be expected in half a minute, and much less than what would be generated over the course of many hours.

The RMS error for the X-coordinates ranges from 7mm to 14mm and the RMS error for the Y-coordinates ranges from 13mm to 24mm.Collecting all the steps together, the RMS errors of the X- and Y-coordinates were found to be 11mm and 17mm respectively.

There are a few contributing factors to the RMS errors. The biggest factor is the positioning of the FSRs. COPP coordinates generated from insole data are limited to positions inside a convex hull of the positions of the FSRs. For example, the COPP will never reach Y-coordinates higher than the center of the FSR under the big toe which, just prior to toe-off, forces a comparison of coordinates inside the hull (insole COPP) to those outside the hull (force plate COPP). Furthermore, this error is exacerbated by the insole COPP veering toward the FSR under the big toe just prior to toe-off. The positions of the FSRs were distributed under key areas of the foot (heel, big toe, first and fifth metatarsal heads), but as can be seen in Fig. 1, no FSR was placed under the small toes. Placing an FSR there may reduce the error by providing a counterpoint to the big toe influence.

Another factor is the size of the FSRs, which have a circular active area with a radius of 7mm. When an FSR output is non-zero, it is assumed via our calculations that the force applied to the FSR is concentrated at the center of the active area. Therefore, a force applied off-center by 5mm will yield a COPP position in error of at least 5mm. Including the fact that the PDMS layer allows some distribution of force around the application point, this error is likely to be on the order of 10mm.

Given these limitations, the similarity shown in Fig. 4 between the force plate COPP curve and the insole COPP curve is striking.

C. Comparison of Spatiotemporal Variables

Previous work examined the difference in stance times found using the insole and a force plate, and the results showed an average difference of -3 ± 29ms [22]. Whereas our results (83 ± 15ms) show a large increase in the bias offset, we have decreased the standard deviation by almost half. Our current results are preferable because a decrease in the standard deviation corresponds to an increase in accuracy (with the understanding that one can always adjust for the bias offset).

Table 2 shows the values of RollSpeed and PathDistance as calculated from the force plate and insole. These variables are of interest because they have the potential to capture different types of steps. A short PathDistance should correspond to abnormal stances such as a flat-footed heel-strike, walking only on the heel, or walking only on the toe, whereas a long PathDistance should correspond to a more typical stance that proceeds from a true heel-strike to a true toe-off. Similarly, a very slow RollSpeed could indicate the same types of steps as a short PathDistance whereas a fast RollSpeed could indicate "floppy" steps wherein there is diminished use of the tibialis anterior to slow plantarflexion.

Unlike the position-by-position analysis possible with COPP coordinates, only one value of these variables is calculated per step. As seen in Table 2, the error in PathDistance and RollSpeed is 28 ± 6% and 19 ± 7% respectively. This error no doubt comes from the errors already discussed in calculating the COPP.

D. Force Plate vs. Instrumented Insole

A force plate and instrumented insole are two very different measurement systems, and it is important to consider the differences when comparing the outputs of these systems. A force plate is made for making highly accurate and highly precise measurements, and an insole is in its construction a very rough approximation of a force plate. Other insole systems exist that exhibit better correlation and lower RMS errors than observed here, but those same systems are exceedingly expensive. Generally the cost of a high-quality force plate will be on the order of at least $10,000 USD, and insole systems with comparable accuracy are similarly priced [14], [15]. One of our insoles costs less than $150 USD in prototype quantities, and our work here has achieved remarkable accuracy given the current limitations. Future work will attempt to increase the COPP through several means such as FSR positioning, individual FSR calibration, and PCB alternatives.

The major advantage of insoles over a force plate is their versatility. Whereas a force plate can be used only in a controlled setting under contrived conditions, insoles allow the subject the freedom to walk anywhere they might normally walk. The insoles also record far more data in far less time than is possible with a force plate. Indeed, the insoles calculate results on every step, not just those that land in a specified area. In addition, the use of on-board microcontrollers and wireless connectivity readily enable the ability to analyze data in real-time, as in [22].

If the home environment is the place people are most likely to experience a fall, then it is important to measure and monitor gait in the home environment. Devices such as our inexpensive instrumented insoles readily facilitate this measurement, and have the potential to provide real-time feedback to prevent falls. Of course, long-term monitoring poses a few challenges regarding the lifetime limitations of FSRs, such as sensor changes due to temperature, drift, and hysteresis. In our scheme, long-term denotes a time frame on the order of a few days, and we will need to determine if sensor changes in this time frame are appreciable enough to affect the consequent analysis.

In addition, many researchers (e.g. [23]) have demonstrated that variability (either the standard deviation or the coefficient of variation calculated by the standard deviation of a measure divided by its mean, such as stride time variability) is key for evaluating fall risk in older adults. These variabilities are likely to be large enough to be readily measured by our instrumented insoles. In [23], the variability was reported as standard deviations: 106 ± 30ms in 20 subjects who had a fall during the subsequent year, and 49 ± 4ms in 32 who did not experience a fall. These magnitudes dwarf the 15ms of standard deviation in the error compared to the force plate. Similarly, we measured greater than a 300% change in COPP area (e.g. as determined using the x and y COPP locations) of a subject with Parkinson's *off* and *on* medication [18]. Though daily changes are likely to be much smaller than changes seen Parkinson's *off* and *on* medication, we expect that we will be able to measure and detect many relevant changes in balance through use of our COPP measures.

V. Conclusion

We have demonstrated that an insole embedded with FSRs can be remarkably accurate in calculating COPP when compared with the current gold-standard of a force plate. In addition, the insole is also accurate in calculating variables (derived from COPP) that are potential indicators of abnormal gait. Given the orders of magnitude of difference in cost and versatility in use for long-term monitoring, instrumented insoles are well-suited for gait analysis when a force plate proves impractical.

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