Journal of Biomechanics 61 (2017) 193–198

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Validation of a smart shoe for estimating foot progression angle during walking gait

Haisheng Xia [a], Junkai Xu [a], Jianren Wang [a], Michael A. Hunt [b], Peter B. Shull [a,*]

[a] State Key Laboratory of Mechanical System and Vibration, School of Mechanical Engineering, Shanghai Jiao Tong University, Shanghai 200240, China
[b] Department of Physical Therapy, University of British Columbia, Vancouver, BC, Canada

ARTICLE INFO

Article history:
Accepted 16 July 2017

Keywords:
Wearable sensing
Sensorized shoe
Knee osteoarthritis

ABSTRACT

The foot progression angle is an important measurement related to knee loading, pain, and function for individuals with knee osteoarthritis, however current measurement methods require camera-based motion capture or floor-embedded force plates confining foot progression angle assessment to facilities with specialized equipment. This paper presents the validation of a customized smart shoe for estimating foot progression angle during walking. The smart shoe is composed of an electronic module with inertial and magnetometer sensing inserted into the sole of a standard walking shoe. The smart shoe charges wirelessly, and up to 160 h of continuous data (sampled at 100 Hz) can be stored locally on the shoe. For validation testing, fourteen healthy subjects were recruited and performed treadmill walking trials with small, medium, and large toe-in (internal foot rotation), small, medium, and large toe-out (external foot rotation) and normal foot progression angle at self-selected walking speeds. Foot progression angle calculations from the smart shoe were compared with measurements from a standard motion capture system. In general, foot progression angle values from the smart shoe closely followed motion capture values for all walking conditions with an overall average error of 0.1 ± 1.9 deg and an overall average absolute error of 1.7 ± 1.0 deg. There were no significant differences in foot progression angle accuracy across the seven different walking gait patterns. The presented smart shoe could potentially be used for knee osteoarthritis or other clinical applications requiring foot progression angle assessment in community settings or in clinics without specialized motion capture equipment.

1. Introduction

The foot progression angle is an important measurement associated with knee joint loading in knee osteoarthritis patients (Hunt et al., 2008), healthy adults (Andrews et al., 1996), and adolescents (Lin et al., 2001). Changing the foot progression angle can reduce the knee adduction moment (Guo et al., 2007; Rutherford et al., 2008; Shull et al., 2013a; Teichtahl et al., 2006; van den Noort et al., 2013), which is a validated surrogate measure of knee loading (Zhao et al., 2007) and is linked to presence (Hurwitz et al., 2002), severity (Sharma et al., 1998), progression (Chang et al., 2007; Miyazaki et al., 2002) and pain (Thorp et al., 2007) in knee osteoarthritis. Altering the foot progression angle may prevent the knee osteoarthritis progression (Chang et al., 2004). Altering the foot progression angle can also reduce knee pain and improve symptoms for individuals with symptomatic knee osteoarthritis (Hunt and Takacs, 2014; Shull et al., 2013b). However, current approaches for estimating the foot progression angle typically require camera-based motion capture or floor-embedded force plates (Chang et al., 2007; Simic et al., 2013) confining foot progression angle assessment and training to laboratories and clinics with specialized equipment. In addition, some clinical applications necessitate continuous monitoring outside the laboratory, such as long-term gait retraining, which may require foot progression angle estimation throughout the day to determine adherence to a new gait pattern.

One approach to estimating the foot progression angle outside the laboratory could be foot-worn sensors. For example, Sabatini et al. (2005) developed a wearable inertial measurement system for estimating stride time, stride length, walking speed, incline and relative stance, and Barth et al. (2015) used a shoe-mounted inertial sensor to perform individualized stride segmentation during gait. Tien et al. (2010) used a foot-mounted inertial measurement system to estimate 3D displacements and rotations for potentially diagnosing gait-related neurological disorders, and

* Corresponding author at: Room 930, Mechanical Engineering Bld. A, School of Mechanical Engineering, Shanghai Jiao Tong University, 800 Dong Chuan Road, Shanghai 200240, China.

E-mail address: pshull@sjtu.edu.cn (P.B. Shull).

http://dx.doi.org/10.1016/j.jbiomech.2017.07.012
0021-9290/

Rouhani et al. (2014) designed a foot-worn system for estimating multi-segment foot kinetics. Sazonov et al. (2011) designed a shoe-based wearable system including an accelerometer and force sensitive resistors to classify six kinds of human activities, and van den Noort et al. (2012) used instrumented force shoes with inertial and magnetic sensors worn on the shoes to estimate gait kinetics and kinematics.

Additionally, several sensor fusion algorithms have been proposed for estimating gait parameters via foot-worn sensors including an extended Kalman filter for orientation estimation (Sabatini, 2006), a stance phase detector using zero-velocity update (Skog et al., 2010), and a gradient descent algorithm to compensate for gyroscope integration error (Madgwick and Harrison, 2011). Huang et al. (2016) presented a foot-worn sensor algorithm for estimating the foot progression angle based on the difference between the foot vector and the heading vector during stance. While foot-worn sensing makes it possible to estimate the foot progression outside of the laboratory, this approach requires external electronics to be worn on the outside of the shoe which may look strange, be uncomfortable, and often requires extra electronics (e.g. battery and data storage) to be carried on the body elsewhere. This approach may also be prone to estimation inaccuracies because of extraneous sensor movement relative to the shoe particularly during gait phase transitions.

Another possible approach to estimating the foot progression angle outside the laboratory could be shoe-embedded or insole-embedded sensors. Pappas et al. (2004) used gyroscope and force sensitive resistors embedded in a shoe insole for gait phase detection and combined this with functional electrical stimulation as a walking aid for people with a drop foot. Bamberg et al. (2008) developed a wearable system with inertial and other sensing to detect heel-strike, toe-off, foot orientation and foot position. Kong and Tomizuka (2009) developed a gait monitoring system for ground contact forces measurement by embedding air pressure sensors between the cushion pad and the sole. Zhou et al. (2011) used a terrain relative velocity sensor embedded in the heel to detect zero velocity periods during walking for pedestrian navigation. Oh and Gross (2015) embedded an accelerometer, vibration motors and a smart phone in the sole of a shoe to give feedback to patients with dementia after long periods of inactivity and to track their location. Tao et al. (2016) used a force sensor embedded shoe sole for ground reaction force measurement to evaluate a rehabilitation training device to help regain walking ability. Despite research efforts involving embedded shoe and insole electronics, there are currently no validated sensorized shoes for estimating the foot progression angle.

The purpose of this paper was to present the validation of a customized smart shoe for estimating foot progression angle during walking. Such a smart shoe could enable foot progression angle estimation outside of a clinical or laboratory setting, making it more feasible to perform daily gait assessment. This could also enable long-term gait monitoring to ensure individuals are following correct gait patterns after gait retraining or be used as a diagnostic tool for those without access to specialized motion capture laboratories to screen for individuals that might be susceptible to musculoskeletal disease, such as knee osteoarthritis.

2. Methods

2.1. Smart shoe design

The smart shoe was composed of an electronic module inserted into the sole of a standard walking shoe (M18972, LZBU) (Fig. 1). The electronic module contains a microcontroller (STM32, STMicroelectronics), 3-axis accelerometer, 3-axis gyroscope and 3-axis magnetometer (MPU9150, Invensense), microSD card (microSDHC Class4, Transcend), 1000 mA h lithium-ion battery, and a wireless charging receiver (Qi, WPC). To protect the electronic components and wire connections, the entire module was encapsulated in silicon and epoxy. The overall size and mass of the electronic module was 52 mm × 40 mm × 13 mm, and 44 g, respectively. The smart shoe charges wirelessly, and the battery lasts for 16 h of continuous use between charges. Sensor data are sampled at 100 Hz, and up to 160 h of continuous data can be stored on the 8 GB microSD card.

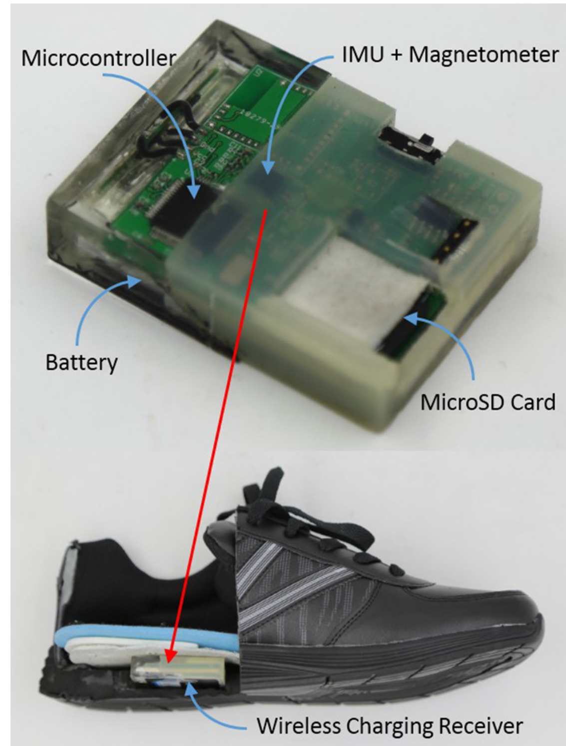

Fig. 1. Smart shoe used to estimate the foot progression angle during walking. (top) Custom electronic module consisting of a microcontroller, 3-axis accelerometer, 3-axis gyroscope, 3-axis magnetometer, 1000 mA h lithium-ion battery, a microSD card and a wireless charging receiver. (bottom) The electronic module is embedded in the sole of a standard walking shoe and can be charged wirelessly.

A foot progression angle estimation algorithm based on inertial and magnetometer sensing (Huang et al., 2016) was programmed onto the electronic module based on orientation estimation, stance phase identification, trajectory estimation, heading vector estimation, and foot vector estimation (Fig. 2). Orientation estimation was computed by integrating angular velocity and corrected with information from the accelerometer and magnetometer, and stance phase identification was performed via zero-velocity detection. The foot vector and heading vector were computed based on the results from orientation and trajectory estimation, respectively, and the foot progression angle was the difference between those two vectors projected in the horizontal plane. For specific details on how to implement this algorithm, see Huang et al. (2016). To avoid the disturbance from the magnetometer, we performed calibration to get an accurate magnetic north estimate when developing the smart shoe module, which followed the calibration procedure described in (Vasconcelos et al., 2011).

2.2. Experimental validation testing

To quantify the accuracy of the smart shoe for measuring foot progression angle, fourteen healthy male subjects (age

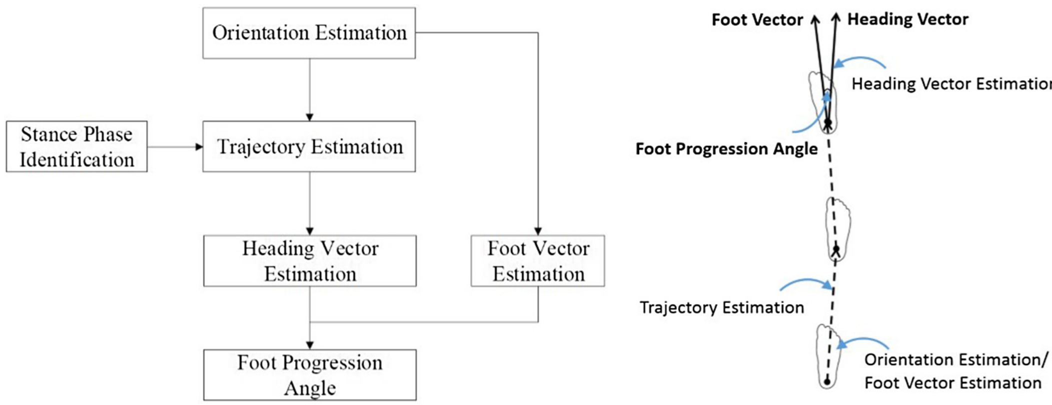

Fig. 2. Foot progression angle estimation algorithm. (left) Orientation estimation is computed by integrating angular velocity and corrected with information from the accelerometer and magnetometer, and stance phase identification is performed via zero-velocity detection. The foot vector and heading vector are computed based on the results from orientation and trajectory estimation, respectively. (right) Foot progression angle is computed as the difference between the heading vector and foot vector projected in the horizontal plane. This figure was modified and used with permission from Huang et al. (2016).

25.4 ± 3.2 years, height 1.73 ± 0.08 m, mass 63.7 ± 8.5 kg, foot size 41.7 ± 1.4 EUR) participated in this study after giving informed consent in accordance with the Declaration of Helsinki. Each subject wore a pair of customized smart walking shoes (five shoe sizes available from 40 to 44 EUR) as described above. Raw sensor data from the smart shoe and motion capture data (Vicon, Oxford, UK) were collected simultaneously at 100 Hz as subjects walked on a treadmill (Bertec, Ohio, USA). Reflective markers were placed on the head of the second metatarsal and the calcaneus to form the foot vector based on motion analysis data, and the difference between the line connecting these markers and the direction of forward progress (defined to align with the long edge of the treadmill pointing in the direction subjects face while walking) was defined as the foot progression angle (Shull et al., 2013b). Unlike foot-worn sensors mounted on top of the shoe which may be prone differing degrees of misalignment each time the sensor is attached and thus require calibration (Dadashi et al., 2013; Huang et al., 2016; Mariani et al., 2012), the smart shoe does not require calibration, since the electronic module is embedded in the sole of the smart shoe. Instead, the offset between the forward horizontal axis of the sensor and the foot vector formed via reflective markers placed over the calcaneus and head of the second metatarsal was calculated for each subject (Fig. 3).

Participants performed walking trials for different self-selected foot progression angles including both toe-in (internal foot rotation) and toe-out (external foot rotation) (Fig. 4). Prior to the formal test, subjects were told to try different foot progression angles and were instructed that the large foot progression angles would be the largest self-selected toe-in and toe-out angles that they could comfortably maintain, and the medium and small foot progression angles should be half and a quarter as much change as the largest toe-in and toe-out angles. Altogether the following seven walking conditions were performed: large toe-in, medium toe-in, small toe-in, normal, small toe-in, medium toe-in and large toe-out, and the trial order was randomized for each subject. Subjects walked at a self-selected speed (1.16 ± 0.06 m/s) which remained the same across all seven walking conditions, for each individual and each trial lasted 2 min. After each trial, subjects were given the option to rest for 1 min or longer if desired.

Fig. 3. Sensor alignment offset computed before testing as the angle between the forward horizontal axis of the sensor and a straight line between a marker placed over the head of the second metatarsal and a marker placed over the calcaneus (Shull et al., 2013b).

2.3. Data analysis

Foot progression angle was defined in the laboratory horizontal plane as the angle between the line connecting the calcaneus and second metatarsal head and the line of forward progression and was computed as the average from 20% to 80% of stance (Huang et al., 2016). Toe-out was considered positive. The error in foot progression angle estimation of the smart shoe for each step was calculated as the difference in foot progression angle estimation between the smart shoe and motion capture system less the sensor alignment offset (Fig. 3). For each trial, the 50 steps preceding the final 20 steps were analyzed to calculate foot progression angle errors. Data analyses were performed on each subject's left foot.

A two-way, random-effects, single measure intraclass correlation coefficients ($ICC_{(2,1)}$) model was used to assess absolute agreement and point estimates of the ICCs were interpreted as follows: excellent (0.75–1), modest (0.4–0.74), or poor (0–0.39) (Clark et al., 2010). Paired Student's t-tests were used for comparing smart shoe and motion capture foot progression angle estimations and Bonferroni correction was used to account for multiple comparisons. One-way analysis of variance (ANOVA) was used to determine if there was any difference in foot progression angle estimation error among the seven different walking gait patterns, and in the case

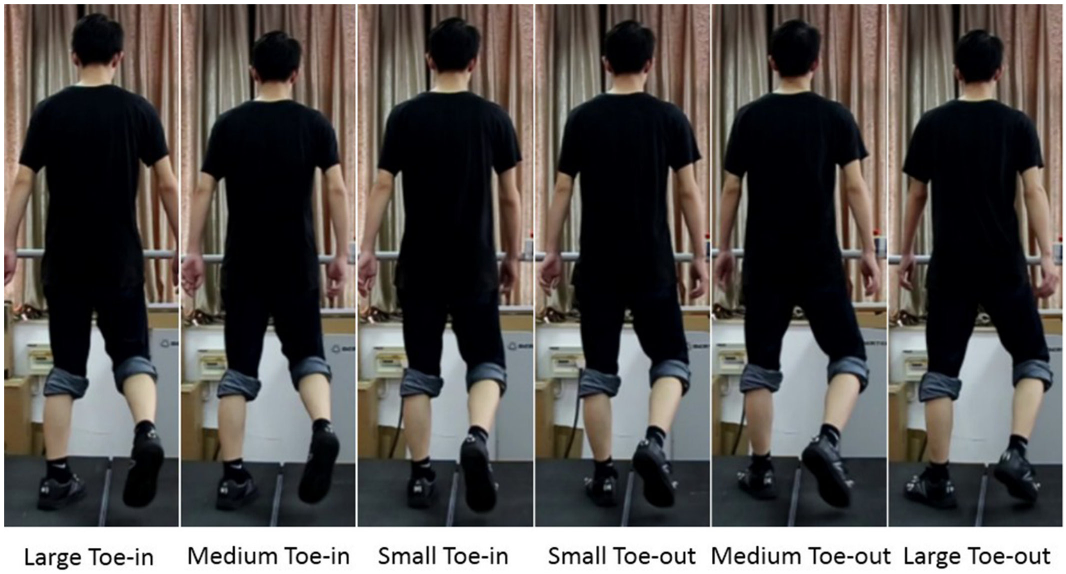

Fig. 4. Example smart shoe treadmill validation modified foot progression angle walking trials. Participants performed the above six gait patterns with modified foot progression angles and a normal walking trial.

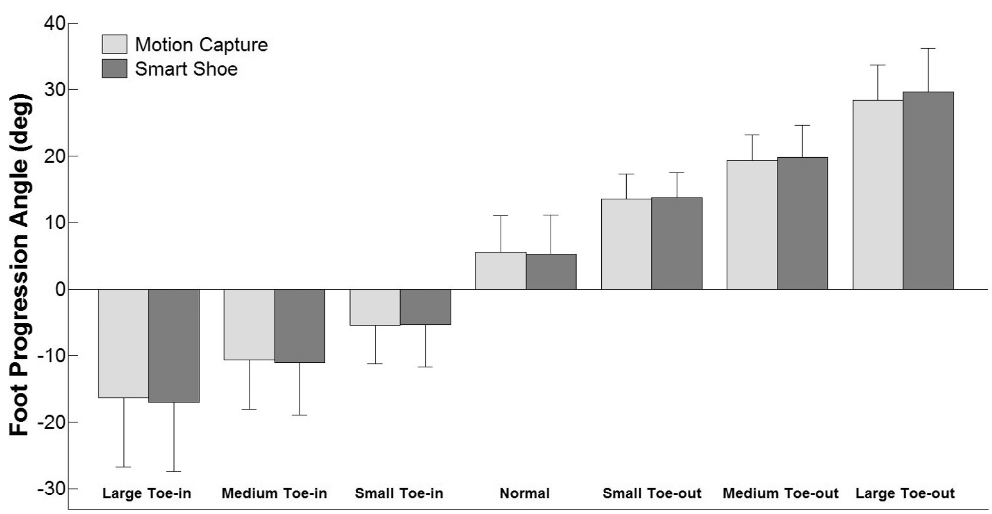

Fig. 5. Foot progression angle estimation grouped by walking pattern. There were no significant differences between smart shoe estimation and motion capture estimation for any of the walking patterns.

when there was a difference, Tukey's procedure was used for post hoc analysis. Statistical significance was set to $p = 0.05$.

3. Results

In general, foot progression angle estimations from the smart shoe closely followed motion capture measurements for all subjects under all walking conditions. Overall average estimation error across all conditions was 0.1 ± 1.9 deg and overall average absolute estimation error across all conditions was 1.7 ± 1.0 deg. Average absolute estimation errors for large toe-in gait, medium toe-in gait, small toe-in gait, normal gait, small toe-in gait, medium toe-in gait and large toe-out gait were 2.1 ± 1.1 deg, 1.6 ± 1.0 deg, 1.7 ± 0.8 deg, 1.6 ± 1.0 deg, 1.5 ± 0.8 deg, 1.7 ± 1.1 deg and 1.8 ± 1.2 deg, respectively. Maximum individual errors across all 7 gaits for each of the 14 subjects were 4.5 deg, −1.9 deg, −3.6 deg, −3.6 deg, −3.8 deg, −4.1 deg, −1.7 deg, 1.8 deg, 4.4 deg, −3.2 deg, 1.7 deg, 2.8 deg, 3.0 deg and 2.1 deg. There were no significant differences between the smart shoe estimation and the motion capture estimation for each of the seven walking conditions (Fig. 5), and there were no significant differences in foot progression angle estimation accuracy among the seven different walking gait patterns ($p = 0.83$). The absolute agreement between the smart shoe and the motion capture system was excellent ($ICC = 0.997$). Foot progression angle estimations based on the smart shoe and motion capture for all individual trials for all subjects and shown in Appendix Fig. 1.

4. Discussion

This paper presented the first sensorized smart shoe for estimating foot progression angle during walking gait. Validation testing demonstrated that the smart shoe was accurate while walking with self-selected normal, toe-in and toe-out gait patterns as compared to the "gold standard" foot progression angle estimation via motion capture. Smart shoe absolute estimation error was less than 2 degrees which is comparable to or slightly better than other wearable systems for estimating gait parameters. Roetenberg et al. (2007) presented a portable magnetic system combined with inertial sensors for human motion tracking and reported orientation estimation errors of about three degrees during gait. Bamberg et al. (2008) developed a shoe-embedded wearable system for estimating foot pitch and reported mean RMS errors of roughly 5 degrees. Rouhani et al. (2014) presented a foot-worn system for estimating multi-segment foot kinetics including dorsiflexion/plantarflexion, inversion/eversion and internal/external rotation with RMS errors of about one degree. Li et al. (2016) used inertial sensors and magnetometers attached to the body and force sensors embedded in the shoe sole to analyze ankle, knee and hip joint angles and reported normalized RMS errors of 5–8 degrees.

While the average *absolute* foot progression estimation error indicates the expected error on any given step, the average foot progression estimation error indicates the expected error for a group of steps. In some applications, the average error may be more useful than the average absolute error. For example, a clinician may be less interested in knowing what a patient's foot progression angle was on a specific step but instead want to know the patient's average foot progression angle for an entire day or week. The smart shoe is particularly well-suited for these types of applications as the average foot progression estimation error was almost negligible (0.1 degree), which aligns with findings of a previous foot-worn sensor study using the same sensor fusion algorithm (Huang et al., 2016).

While most previous wearable solutions for estimating foot-related gait parameters involve foot-worn sensors, for the presented smart shoe, sensors are embedded inside the shoe. One benefit of sensor-embedded shoes is that it doesn't need to be continually calibrated due to sensor position shifting and misalignment. For example, the foot-worn sensor presented in (Huang et al., 2016) requires calibration trials involving keeping the body stationary and pointing the foot forward for ten seconds to find the vertical axis and then dorsiflexing and plantarflexing three times to find the transverse axis. Foot-worn sensors may be prone to noisier signals because of the extraneous sensor movement relative to the shoe especially during heel strike and toe-off gait phase transitions. In addition, sensor-embedded shoes appear as normal walking shoes while foot-worn sensor shoes may look abnormal making patients less likely to wear them consistently in daily life. In this study, five pairs of shoes were used with the size ranging from 40 to 44 EUR to satisfy foot sizes of the subjects in this study. Because the sensor is implanted in the sole of the shoe (Fig. 1), it could easily be implementable in different shoes of various sizes in future use. A potential disadvantage of sensors embedded in the shoe sole or insole is that the sensor has to withstand large vertical impact forces during stance and also, depending on the sensor location, bending forces during heel strike and toe-off. This requires a more robust sensor packaging design to withstand these large forces as compared to the relatively small impact forces encountered by a foot-worn sensor attached to the top of the shoe.

The validated smart shoe could help to facilitate gait assessment and retraining as a conservative treatment option for early-stage knee osteoarthritis. Guo et al. (2007) studied the influence of foot progression angle on knee osteoarthritis patients during walking and stair climbing and suggested that altering the foot progression angle could benefit individuals with early-stage knee osteoarthritis. More recent studies have shown that weekly gait retraining sessions to alter the foot progression angle can reduce knee loading, reduce knee pain and improved symptoms for individuals with early-stage knee osteoarthritis (Hunt and Takacs, 2014; Shull et al., 2013b). Despite research showing that changing the foot progression angle could be a promising non-surgical treatment for knee osteoarthritis, currently it is only possible to assess foot progression angle inside the laboratory. This makes it difficult to track user compliance of gait modifications in daily life outside of the laboratory, and limits the potential benefits to individuals with access to motion capture facilities. Thus, the validated smart shoe could help to enable more widespread adoption of conservative, non-surgical treatments for knee osteoarthritis.

In addition to knee osteoarthritis, accurate estimation of the foot progression angle outside the laboratory may benefit other clinical applications. Changes in the foot progression angle have been correlated with changes in ankle inversion moment (Andrews et al., 1996), hip joint moment (Bowsher and Vaughan, 1995), foot pressure distribution (Chang et al., 2004; Lai et al., 2014), and foot medial loading (Hastings et al., 2010), and thus it is possible that tracking foot progression in daily life could provide insights into diseases associated with these gait parameters such as hip osteoarthritis (Mont et al., 2007), pronated diabetic foot (Albert and Rinoie, 1993) and peripheral neuropathy (Hastings et al., 2010). The foot progression angle is also an important clinical measurement for assessing patients with clubfoot (Yngve, 1990) and distal tibial physeal fractures (Phan et al., 2002).

One limitation of this study is that we only tested healthy participants with normal gait patterns, thus the accuracy of the smart shoe for movement disorders involving abnormal gaits like foot drag or indistinct heel strikes remain unknown. However, our aim was simply to validate the mechanism and algorithms of the sensor against a gold standard, and this comparison should be independent of the user. In this study, we did not observe any dramatic changes in the raw magnetometer readings or FPA estimates that would be expected from significant magnetic disturbances, however, caution should be taken in future testing as the magnetometer could still be susceptible to relatively large unaccounted for magnetic fields. Another limitation of this study is that we do not have over-ground walking data to validate the smart shoe, and it is possible that the accuracy of treadmill and over-ground walking may differ. Also, walking trials were only performed at subjects' self-selected speeds, thus it is possible that accuracy results could differ for significantly slower walking or faster walking and running gaits. More research is needed to explore smart shoe accuracy for different gait speeds, different walking environment such as over-ground walking, and abnormal gait patterns such as foot drag or indistinct heel strikes. Additionally, in this study the primary focus was on quantifying the overall accuracy of the foot progression angle estimate. Because the overall foot progression angle errors were relatively low, we did not perform additional analysis to quantify the specific accuracies of the foot and heading vectors though future research could focus on this to potentially reveal additional insights.

In conclusion, this paper presented a smart shoe for estimating foot progression angle. The smart shoe could potentially be used for knee osteoarthritis or other clinical applications requiring foot progression angle assessment in daily life or in clinics without specialized motion capture equipment. Future versions of the smart shoe could potentially add feedback functionality through a wireless connected smart phone or embedded vibration devices in the shoe for real-time gait retraining to enable individuals to modify their gait outside of traditional laboratory settings.

Acknowledgments

This work was supported by the National Natural Science Foundation of China (51505283), the University of Michigan–Shanghai Jiao Tong University Collaborative Research in Applications of Nanotechnology Fund (14X120010006), the Science and Technology Commission of Shanghai Municipality, Yangfan Program (14YF1401600), and the Natural Sciences and Engineering Research Council of Canada.

Conflict of interest

None of the authors had any conflict of interest regarding this manuscript.

Contributing authors

All authors have made substantial contributions to the following: (1) the conception and design of the study, or acquisition of data, or analysis and interpretation of data, (2) drafting the article or revising it critically for important intellectual content, (3) final approval of the version to be submitted. Each of the authors has read and concurs with the content in the manuscript. The manuscript and the material within have not been and will not be submitted for publication elsewhere.

Appendix A. Supplementary materials

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jbiomech.2017.07.012.

References

Albert, S., Rinoie, C., 1993. Effect of custom orthotics on plantar pressure distribution in the pronated diabetic foot. J. Foot Ankle Surg. 33, 598–604.
Andrews, M., Noyes, F.R., Hewett, T.E., Andriacchi, T.P., 1996. Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: A critical analysis of the reliability of gait analysis data. J. Orthop. Res. 14, 289–295.
Bamberg, S., Benbasat, A.Y., Scarborough, D.M., Krebs, D.E., Paradiso, J.A., 2008. Gait analysis using a shoe-integrated wireless sensor system. IEEE Trans. Inf. Technol. Biomed. 12, 413–423.
Barth, J., Oberndorfer, C., Pasluosta, C., Schülein, S., Gassner, H., Reinfelder, S., Kugler, P., Schuldhaus, D., Winkler, J., Klucken, J., Eskofier, B.M., 2015. Stride segmentation during free walk movements using multi-dimensional subsequence dynamic time warping on inertial sensor data. Sensors (Switzerland) 15, pp. 6419–6440.
Bowsher, K., Vaughan, C., 1995. Effect of foot-progression angle on hip joint moments during gait. J. Biomech. 28, 759–762.
Chang, A., Hurwitz, D., Hayes, K., Dunlop, D., Song, J., Genge, R., Cahue, S., Sharma, L., 2004. Toe-out angle during gait and protection against progression of medial tibiofemoral knee. Arthritis Rheum. 50, S252–S253.
Chang, A., Hurwitz, D., Dunlop, D., Song, J., Cahue, S., Hayes, K., Sharma, L., 2007. The relationship between toe-out angle during gait and progression of medial tibiofemoral osteoarthritis. Ann. Rheum. Dis. 66, 1271–1275.
Chang, W., Tsirikos, A., Miller, F., Schuyler, J., Glutting, J., 2004. Impact of changing foot progression angle on foot pressure measurement in children with neuromuscular diseases. Gait Posture 20, 14–19.
Clark, R.A., Bryant, A.L., Pua, Y., McCrory, P., Bennell, K., Hunt, M., 2010. Validity and reliability of the Nintendo Wii Balance Board for assessment of standing balance. Gait Posture 31, 307–310.
Dadashi, F., Mariani, B., Rochat, S., Büla, C.J., Santos-Eggimann, B., Aminian, K., 2013. Gait and foot clearance parameters obtained using shoe-worn inertial sensors in a large-population sample of older adults. Sensors (Basel). 14, 443–457.
Guo, M., Axe, M.J., Manal, K., 2007. The influence of foot progression angle on the knee adduction moment during walking and stair climbing in pain free individuals with knee osteoarthritis. Gait Posture 26, 436–441.
Hastings, M.K., Gelber, J.R., Isaac, E.J., Bohnert, K.L., Strube, M.J., Sinacore, D.R., 2010. Foot progression angle and medial loading in individuals with diabetes mellitus, peripheral neuropathy, and a foot ulcer. Gait Posture 32, 237–241.
Huang, Y., Jirattigalachote, W., Cutkosky, M., Zhu, X., Shull, P., 2016. Novel foot progression angle algorithm estimation via foot-worn, magneto-inertial sensing. IEEE Trans. Biomed. Eng. 63, 2278–2285.
Hunt, M.A., Birmingham, T.B., Bryant, D., Jones, I., Giffin, J.R., Jenkyn, T.R., Vandervoort, A.A., 2008. Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. Osteoarthr. Cartil. 16, 591–599.
Hunt, M.A., Takacs, J., 2014. Effects of a 10-week toe-out gait modification intervention in people with medial knee osteoarthritis: A pilot, feasibility study. Osteoarthr. Cartil. 22, 904–911.
Hurwitz, D.E., Ryals, A.B., Case, J.P., Block, J.A., Andriacchi, T.P., 2002. The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. J. Orthop. Res. 20, 101–107.
Kong, K., Tomizuka, M., 2009. A gait monitoring system based on air pressure sensors embedded in a shoe. IEEE/ASME Trans. Mechatron. 14, 358–370.
Lai, Y.C., Lin, H.S., Pan, H.F., Chang, W.N., Hsu, C.J., Renn, J.H., 2014. Impact of foot progression angle on the distribution of plantar pressure in normal children. Clin. Biomech. 29, 196–200.
Li, G., Liu, T., Yi, J., Wang, H., Li, J., Inoue, Y., 2016. The lower limbs kinematics analysis by wearable sensor shoes. IEEE Sens. J. 16, 2627–2638.
Lin, C.J., Lai, K.A., Chou, Y.L., Ho, C.S., 2001. The effect of changing the foot progression angle on the knee adduction moment in normal teenagers. Gait Posture 14, 85–91.
Madgwick, S., Harrison, A., 2011. Estimation of IMU and MARG orientation using a gradient descent algorithm. 2011 IEEE Int. Conf. Rehabil. Robot, pp. 1–7.
Mariani, B., Rochat, S., Büla, C.J., Aminian, K., 2012. Heel and toe clearance estimation for gait analysis using wireless inertial sensors. IEEE Trans. Biomed. Eng. 59, 3162–3168.
Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., Shimada, S., 2002. Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Ann. Rheum. Dis. 61, 617–622.
Mont, M.A., Seyler, T.M., Ragland, P.S., Starr, R., Erhart, J., Bhave, A., 2007. Gait analysis of patients with resurfacing hip arthroplasty compared with hip osteoarthritis and standard total hip arthroplasty. J. Arthroplasty 22, 100–108.
Oh, H., Gross, M.D., 2015. Awareable steps: functional and fashionable shoes for patients with dementia. In: UbiComp '15 Proc. 2015 ACM Int. Jt. Conf. Pervasive Ubiquitous Comput, pp. 579–583.
Pappas, I., Keller, T., Mangold, S., 2004. A reliable gyroscope-based gait-phase detection sensor embedded in a shoe insole. IEEE Sens. 4, 268–274.
Phan, V., Wroten, E., Yngve, D., 2002. Foot progression angle after distal tibial physeal fractures. J. Pediatr. Orthop. 22, 31–35.
Roetenberg, D., Slycke, P.J., Veltink, P.H., 2007. Ambulatory position and orientation tracking fusing magnetic and inertial sensing. IEEE Trans. Biomed. Eng. 54, 883–890.
Rouhani, H., Favre, J., Crevoisier, X., Aminian, K., 2014. A wearable system for multi-segment foot kinetics measurement. J. Biomech. 47, 1704–1711.
Rutherford, D.J., Hubley-Kozey, C.L., Deluzio, K.J., Stanish, W.D., Dunbar, M., 2008. Foot progression angle and the knee adduction moment: a cross-sectional investigation in knee osteoarthritis. Osteoarthr. Cartil. 16, 883–889.
Sabatini, A., Martelloni, C., Scapellato, S., 2005. Assessment of walking features from foot inertial sensing. IEEE Trans. Biomed. Eng. 52, 486–494.
Sabatini, A.M., 2006. Quaternion-based extended Kalman Filter for determining orientation by inertial and magnetic sensing. IEEE Trans. Biomed. Eng. 53, 1346–1356.
Sazonov, E.S., Fulk, G., Hill, J., Schutz, Y., Browning, R., 2011. Monitoring of posture allocations and activities by a shoe-based wearable sensor. IEEE Trans. Biomed. Eng. 58, 983–990.
Sharma, L., Hurwitz, D.E., Thonar, E.J.M.A., Sum, J.A., Lenz, M.E., Dunlop, D.D., Schnitzer, T.J., Kirwan-Mellis, G., Andriacchi, T.P., 1998. Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis Rheum. 41, 1233–1240.
Shull, P.B., Shultz, R., Silder, A., Dragoo, J.L., Besier, T.F., Cutkosky, M.R., Delp, S.L., 2013a. Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis. J. Biomech. 46, 122–128.
Shull, P.B., Silder, A., Shultz, R., Dragoo, J.L., Besier, T.F., Delp, S.L., Cutkosky, M.R., 2013b. Six-week gait retraining program reduces knee adduction moment, reduces pain, and improves function for individuals with medial compartment knee osteoarthritis. J. Orthop. Res. 31, 1020–1025.
Simic, M., Wrigley, T.V., Hinman, R.S., Hunt, M.A., Bennell, K.L., 2013. Altering foot progression angle in people with medial knee osteoarthritis: The effects of varying toe-in and toe-out angles aremediated by pain and malalignment. Osteoarthr. Cartil. 21, 1272–1280.
Skog, I., Handel, P., Nilsson, J., 2010. Zero-velocity detection—An algorithm evaluation. IEEE Trans. Biomed. Eng. 57, 2657–2666.
Tao, W., Zhang, J., Li, G., Liu, T., Liu, F., 2016. A wearable sensor system for lower-limb rehabilitation evaluation using the GRF and CoP distributions. Meas. Sci. Technol. 27, 1–13.
Teichtahl, A.J., Morris, M.E., Wluka, A.E., Baker, R., Wolfe, R., Davis, S.R., Cicuttini, F. M., 2006. Foot rotation-A potential target to modify the knee adduction moment. J. Sci. Med. Sport 9, 67–71.
Thorp, L.E., Sumner, D.R., Wimmer, M.A., Block, J.A., 2007. Relationship between pain and medial knee joint loading in mild radiographic knee osteoarthritis. Arthritis Care Res. 57, 1254–1260.
Tien, I., Glaser, S.D., Bajcsy, R., Goodin, D.S., Aminoff, M.J., 2010. Results of using a wireless inertial measuring system to quantify gait motions in control subjects. IEEE Trans. Inf. Technol. Biomed. 14, 904–915.
van den Noort, J.C., Schaffers, I., Snijders, J., Harlaar, J., 2013. The effectiveness of voluntary modifications of gait pattern to reduce the knee adduction moment. Hum. Mov. Sci. 32, 412–424.
van den Noort, J.C., van der Esch, M., Steultjens, M.P.M., Dekker, J., Schepers, H.M., Veltink, P.H., Harlaar, J., 2012. The knee adduction moment measured with an instrumented force shoe in patients with knee osteoarthritis. J. Biomech. 45, 281–288.
Vasconcelos, J.F., Elkaim, G., Silvestre, C., Oliveira, P., Cardeira, B., 2011. Geometric approach to strapdown magnetometer calibration in sensor frame. IEEE Trans. Aerosp. Electron. Syst. 47, 1293–1306.
Yngve, D., 1990. Foot-progression angle in clubfeet. J. Pediatr. Orthop. 10, 467–472.
Zhao, D., Banks, S.A., Mitchell, K.H., D'Lima, D.D., Colwell, C.W., Fregly, B.J., 2007. Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. J. Orthop. Res. 25, 789–797.
Zhou, C., Downey, J., Choi, J., Stancil, D., 2011. A shoe-embedded RF sensor for motion detection. IEEE Microw. Theory Tech. Soc. 21, 169–171.