Validation of Temporal Gait Metrics from three IMU Locations to the Gold Standard Force Plate

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Abstract— **The purpose of this work is to compare temporal gait parameters from three different IMU locations to the gold standard force platform. 33 subjects (12 F, 21 M) performed twenty gait trials each while wearing inertial measurement units (IMUs) on the trunk, both shanks and both feet. Data was simultaneously collected from a laboratory embedded force plate. Step times were derived from the raw IMU data at the three IMU locations using methods that have been shown to be accurate. Step times from all locations were valid compared to the force plate. Foot IMU step time was the most accurate (Pearson = .991, CI width = 3.00e2), the trunk IMU was the next most accurate (Pearson = .974, CI width = 4.85e2) and shank step time was the least accurate (Pearson = .958, CI width = 6.80e2). All three sensing locations result in valid estimations of step time compared to the gold standard force plate. These results suggest that the foot location would be most appropriate for clinical applications where very precise temporal parameter detection is required.**

I. Introduction

Recently, there has been an explosion in the use of wearable activity trackers to provide quantitative measures of mobility; such as step counting and sedentary versus non-sedentary time tracking. While such measures can potentially be useful in aiding general weight loss and providing motivation for users to become more active, there is also a need to determine more detailed quality of mobility metrics [1]. Research in the mobility space has linked such detailed mobility metrics to important health related outcomes such as falls risk [2], Parkinson's disease [1, 3], Multiple sclerosis score [4] and cerebral palsy gait assessment [5].

To obtain accurate mobility metrics, a more robust mounting location is required than is used traditionally for consumer wearables. Basic step counting does not provide sufficient detail for understanding the quality of a patients mobility, thus more detailed metrics such as step time, stride time variability and step distance are required. Very detailed gait metrics include kinetic and kinematic measures which are obtained in biomechanics labs, or else remotely through the use of an inertial measurement unit (IMU) on each body segment [6], such systems are beyond the scope of this work. Three of the most commonly used IMU mounting locations to obtain quality of gait metrics in the literature include, an IMU over the spine on the lower back, two IMUs on the lower legs or two IMUs on the dorsal aspect of each foot. A brief summary of research into all three mounting locations is provided below.

Two standard gait analysis techniques involve the use of a trunk IMU. Zjilstra & Hof [7] showed that initial contacts (IC) could be found in the anterior-posterior acceleration signal at the peak value prior to a sharp decrease, which includes a zero-crossing. Many subsequent IC detection methods have been based on this algorithm [8-10]. Alternatively, Moe-Nilssen & Helbostad [11], showed that an unbiased autocorrelation coefficient could be used to detect step and stride times as well as regularity and symmetry metrics. Many subsequent studies have used this method as well [12-14].

Sagittal plane gyroscope data from the shank has been a popular choice of signal to detect gait events [15, 16]. The swing phase of each step is denoted by a large increase in sagittal plane rotation, with IC shown to be located at the trough after this large increase [17, 18]. Recent research has suggested that toe-off (TO) occurs after the gyroscope trough prior to the large increase resulting from the swing phase, at a mid-point between the trough and the non-high pass filtered rotation rates zero-crossing [19, 20].

An IMU on the foot has been shown to be an accurate sensing location for the estimation of step distance [21, 22]. This is because the zero movement during ground contact can be used to reduce integration times to result in more accurate distance estimations [21]. Recent research has also shown how detailed gait events can be found using an inertial sensor on the heel and on the toe [23].

No work has compared all three mounting locations on the same data-set to determine which is the most accurate compared to gold standard motion analysis methods. Previous work has compared temporal features from a shank inertial sensor to temporal features from a trunk inertial sensor and found that the shank sensor was more accurate [24]. In this study, they did not include a comparison to a foot inertial sensor. A limitation of their work may involve their use of treadmill walking, which has been shown to cause different motor patterns than over ground walking [25], as well as introduce different acceleration parameters to the inertial sensors due to the fact the person is not moving across the ground.

The purpose of the current work is to derive step times during over-ground walking from three different IMU locations to determine how they compare to the gold standard force plate as well as how they compare to each other on the same data-set. Since forces get attenuated by muscles and

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joints as they move up the body it is expected that the foot sensor will provide the most accurate estimation, followed by the shank, and finally the lumbar mounting location.

II. METHODS

A. Subjects

33 subjects, 12 female and 21 male (25 years ± 8, height 176cm ± 9, weight 73kg ± 13, BMI 23 ± 3) were recruited from the University campus and wider community through the means of posters and advertisements. Subjects were eligible to participate in this study if they were aged 18+ and were capable of providing informed consent. Exclusion criteria included; (1) unexplained falls within the last year, (2) active medical treatment, (3) fractures, surgery or hospitalization within the last 3 months and (4) serious neurological pathology. Ethical approval was sought and obtained from the University Health Research and Ethics Committee. All subjects provided written informed consent prior to being included.

B. Protocol

The testing protocol consisted of controlled gait trials conducted in the University's motion capture laboratory. The controlled gait trial was designed to allow the direct comparison of step time measures derived from a gold standard force platform and those derived from three algorithms [7, 20, 26]. Each subject's height and weight was obtained. During testing, subjects were instructed to walk along an approximately 10 metre walkway at a self-selected "normal" walking speed. Subjects were asked to focus on a preselected point at the end of the walkway and were not aware of the force platform location to ensure a normal gait pattern. The tester deemed the trial successful if two consecutive force plate heel-strikes were present. The starting leg, force plate steps and the exact walking distance was recorded for each walking trial. The procedure was repeated until 20 successful (10 right foot and 10 left foot force plate steps) trials were completed.

C. Sensor Setup

Five IMUs were used in total; one placed on the dorsal aspect of each foot, one placed on each of the subject's shanks, and one on the trunk. The foot IMUs were placed on the dorsum of each foot so that the distal aspect of the sensor lined up with a perpendicular line coming from the 5th metatarsal. The shank IMUs were placed 10cm superior to the bisection of the lateral malleolus bilaterally. This location ensured minimal soft tissue attachment in order to limit the amount of skin and muscle movement. The lumbar IMU was placed at the level of the 3^{rd} lumbar vertebra in order to closely match the centre of mass acceleration during gait [27, 28]. The foot IMUs were attached using athletic tape, the trunk and shank IMUs were fixed in place with elastic straps and secured using double sided tape, thus reducing any additional sensor movement during gait. Each IMU was time synchronized and the data was stored on board.

D. Force Platform

As a validation tool, two embedded force platforms (AMTI, Watertown, Massachusetts) were used to obtain a gold standard measurement IC. CODA Analysis software V 6.79.3-CX1 (Leicester, UK) was used to record and store data from the force platforms. The force platforms were located in the centre of the walkway. Force platform data was acquired at 1000Hz and was passed through a fourth-order zero phase Butterworth low-pass digital filter with a 6 Hz cut-off frequency [29]. In compliance with the recommendations outlined by Tirosh and Sparrow [30], a vertical force threshold of 10 N was selected to identify initial contact. Step time was calculated at the time from one IC to the next contra-lateral leg IC.

E. IMU Processing

Shimmer3 IMUs (Shimmer, Dublin, Ireland) were used to measure acceleration and angular rate at a sampling rate of 256Hz. Firmware and configuration settings were set using Consensys software (Shimmer, Dublin, Ireland). The tri-axial accelerometer and gyroscope signals were enabled and set to ranges of ±4G and ±1,000 deg/sec, respectively. MATLAB 2014b (Mathworks, Natick, MA, USA) was used to replicate previously published algorithms which were designed to estimate gait events from IMU data.

The algorithm to estimate gait events from the lumbar mounted IMU was based on the seminal paper by Zijlstra & Hof [7], which has been shown to be an accurate estimation of gait events [8]. The algorithm to estimate IC and TO from the shank sensors was based on the use of the sagittal plane gyroscope signal. This algorithm was chosen based on previous work which indicated it was the most accurate algorithm compared to other commonly used shank-based algorithms [16]. The algorithm chosen to estimate step time from the foot sensors was based on the algorithm by Jasiewicz and colleagues [26], which was shown to be accurate compared to a force plate and a shank inertial sensor.

F. Statistics

Statistical analysis was performed in order to determine the validity of the algorithms' estimation of step time. From each laboratory walking trial, four step time values were obtained: one from the gold standard force plate, one from the foot IMUs, one from the shank IMUs and one from the lumbar IMU. Each step time was averaged over all twenty trials for each subject. These average values were used for the final comparison [31]. Pearson correlation coefficient [32] was calculated for both mounting locations to determine the level of correlation between the force plate and the IMU temporal measures. A Bland-Altman style analysis was conducted in order to determine the levels of agreement between the gold standard gait parameters, as determined by the force platform, and those derived from the three gait detection algorithms [31]. Mean difference (MD) between the each of the three estimated step times and the force plate step time was calculated. The 95% confidence interval (CI) was calculated to demonstrate the precision of the estimated limits of agreement, using the formula:

$$CI = X \pm t_{1-a/2}\sqrt{2SD^2} \quad (1)$$

Where X is the mean of the mean differences between the IMU and force plate step time measurements, $\sqrt{2SD^2}$ is the estimated standard deviation, $t_{1-a/2}$ is dependent on the probability level chosen and degrees of freedom [33]. Some of the effects of repeated measurement error have been removed due to the fact that repeated measures on each subject were averaged and the mean values were compared. For this reason the SD of the means was corrected according to methods proposed by Bland & Altman [31]. The CI width was determined in order to estimate the range within which 95% of results should be expected to fall. Confidence interval width percentage is also reported, which is the 95% confidence interval width divided by the average step time from the force plate.

TABLE I. MEAN, MAXIMUM, MINIMUM AND SD (STANDARD DEVIATION) STEP TIME VALUES OBTAINED FROM THE FORCE PLATFORM AND THE THREE IMU MOUNTING LOCATIONS

	Force plate (sec)	*Trunk IMU (sec)*	*Shank IMUs (sec)*	*Foot IMUs (sec)*
Mean	.534	.541	.541	.544
Max	.593	.595	.624	.611
Min	.469	.472	.472	.479
SD	.039	.038	.043	.040

III. RESULTS

A total of 660 complete walking trials, each consisting of a single step, were obtained from 33 participants during the data collection phase of this study. 39 individual steps, dispersed throughout the subjects, had to be excluded from the analysis, due to data corruption at the data collection phase, resulting in a total of N = 621 steps for analysis. Table 1 lists the mean, maximum, minimum and standard deviation (SD) of all averaged step times, as derived from the foot IMUs, the shank IMUs, lumbar IMU and the force platform.

TABLE II. COMPARISON BETWEEN EACH FO THE IMU MOUNTING LOCATIONS AND THE GOLD STANDARD FORCE PLATFORM.

	Lumbar IMU vs FP	*Shank IMUs vs FP*	*Foot IMUs vs FP*
Mean difference	6.25e-3	6.15e-3	-9.20e-3
Corrected SD difference	1.24e-2	1.73e-2	7.66e-3
Pearson product correlation	0.974	0.958	0.991
Upper 95% CI	3.05e-2	4.01e-2	5.84e-3
Lower 95% CI	-1.80e-2	-2.78e-2	-2.42e-2
CI width	4.85e2	6.80e2	3.00e2
CI width % of avg FP step time	9.09	12.72	5.62

SD - standard deviation, CI - confidence interval, avg - average & FP - force plate

Table 2 presents a direct comparison between the estimated step times from each of the IMU locations and the gold standard force platform. The Pearson correlation coefficients show the high levels of correlation between the force plate and IMU step time measures [34]. Additionally, the mean difference between the step times measured from each of the IMU locations and those measured by the force platform is also shown, with step time derived from the shanks showing the smallest mean difference, followed by the lumbar sensor and finally the foot sensor with the largest. The comparison between the step times estimated from each of the IMU locations and from the force platform have been visualized using Bland-Altman style plots, as shown in Figures 1, 2 and 3. Upper and lower 95% CI were found to be relatively small for the foot IMUs and larger for the trunk IMU and the shank IMUs (Table 2). Figure 3 shows a scatter plot illustrating the agreement between the IMU and force platform.

IV. DISCUSSION

The purpose of this paper was to compare how well step time estimated from three different IMU locations compared to step time from the gold standard force plate. The results show that all three locations can accurately estimate step time when compared to the gold standard force plate. However, even though all locations could accurately estimate step time, it was found that step time from the foot IMUs demonstrated the highest level of accuracy, followed by the lumbar IMU and finally shank IMUs. Despite the varying levels of accuracy, each mounting location provided a valid measure of step time.

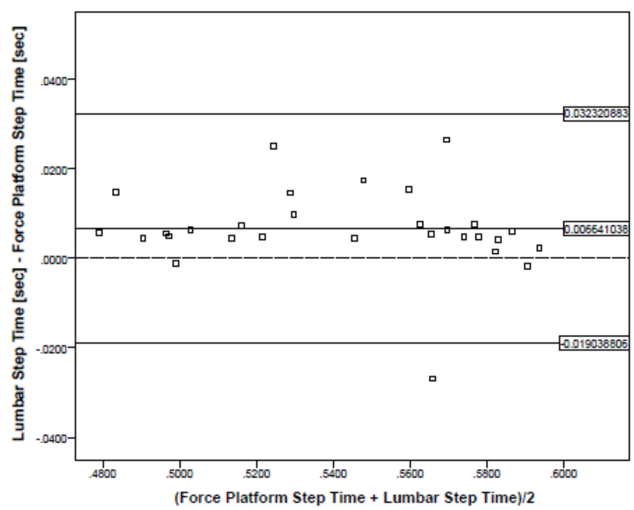

Figure 1. Bland-Altman plot comparing step time from the force plate and the trunk IMU.

Recent work has shown a Pearson product correlation of 0.997 between step time from a lumbar IMU and a GaitRite mat [8]. In the current work we used the same processing methods as Godfrey and colleagues [8] to estimate step time and found a Pearson product correlation of 0.969 compared to the force plate. A possible limitation of Godfrey et al work is that they did not use the gold standard force plate for comparison.

We hypothesized that as the IMU mounting location moved away from the ground, the estimation of step time would become less accurate due to the attenuation of ground reaction forces up the body. This was not the case, as the shank location had the lowest Pearson product correlation as well as the largest limits of agreement. The shank IMU confidence interval width was over 13% of the average step time from the force plate (Table 2). This means that the shank IMU cannot accurately detect temporal gait differences that are less than 13%.

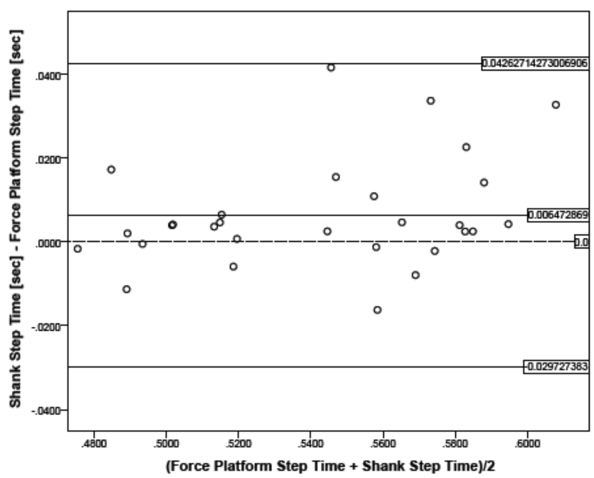

Figure 2. Bland-Altman plot comparing step time from the force plate and the shank IMUs.

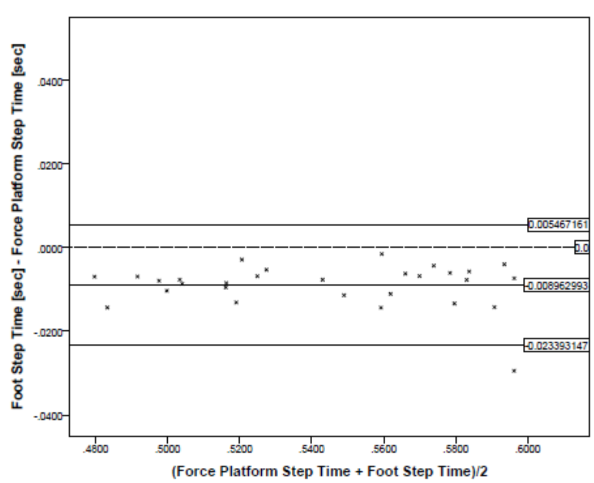

Figure 3. Bland-Altman plot comparing step time from the force plate and the foot IMUs.

The foot IMUs however, were the most accurate at estimating step time compared to the gold standard force plate with a Pearson product correlation of 0.991 and a confidence interval width of 5.6% of the average step time of the force plate (Table 2). The foot inertial sensing location is likely the most accurate because the sensors are closer to the ground than the shank or lumbar IMUs. By their very definition, gait events are defined by ground reaction force changes under foot. Thus, the IMUs on the foot have the best chance of detecting gait events because ground reaction forces will be attenuated by joints and muscles as they move up the body [35].

An important aspect to consider when implementing ambulatory monitoring systems is an individual's comfort and preference. Some individuals may prefer to wear a belt-like lumbar IMU for prolonged periods of time, while others may find it less invasive to wear shank or foot mounted IMUs. Clothing preferences may also be a factor in driving which IMU location a person might prefer. The high level of accuracy obtained from all three IMU locations demonstrates that it may be possible to take into account an individual's preference without adversely affecting accuracy. This is an important consideration, as the wearer's comfort and preference may lead to higher levels of compliance.

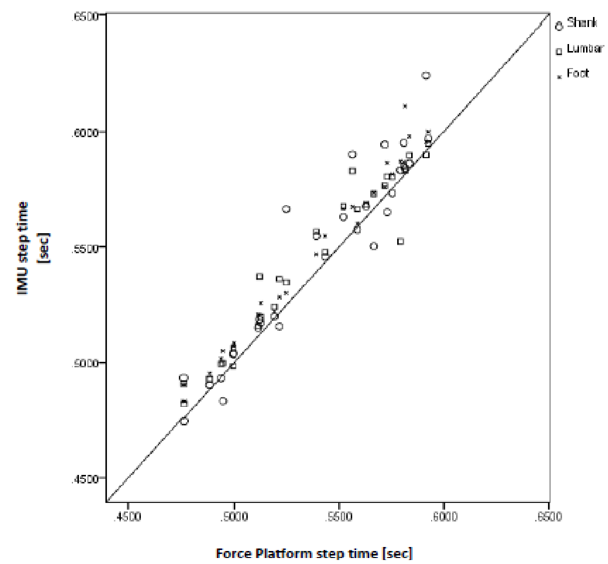

Figure 4. The degree of agreement between the three IMU locations and the force plate. X-axis is step time from the force plate, y-axis is step time from the various IMU locations. The solid line represents where all points would lie if agreement between IMUs and FP were exactly the same.

There were a number of limitations to this study. Firstly, the participants gait speed was not controlled and therefore, it was not possible to assess the validity of the event detection system across a range of different controlled walking speeds. However, the use of the participant's chosen walking speed ensured that individuals were allowed to walk in a natural manner. Secondly, due to the layout of the in-floor force platform system, it was not possible to obtain stride measurements as the force platform was too short to capture two successive ICs on the same leg. This meant that it was not possible to validate the system's capability of accurately estimating stride time.

In this work we have shown that three different inertial sensor mounting locations can be used to obtain valid temporal gait measures as compared to the gold standard force plate. These results suggest that foot location would be the most appropriate for clinical applications where very precise temporal parameter detection is required.

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REFERENCES

[1] A. Weiss, M. Brozgol, M. Dorfman, T. Herman, S. Shema, N. Giladi, and J. M. Hausdorff, "Does the evaluation of gait quality during daily life provide insight into fall risk? A novel approach using 3-day accelerometer recordings," *Neurorehabil Neural Repair,* vol. 27, no. 8, pp. 742-52, Oct, 2013.

[2] J. M. Hausdorff, D. A. Rios, and H. K. Edelberg, "Gait variability and fall risk in community-living older adults: a 1-year prospective study," *Arch Phys Med Rehabil,* vol. 82, no. 8, pp. 1050-6, Aug, 2001.

[3] A. Weiss, S. Sharifi, M. Plotnik, J. P. van Vugt, N. Giladi, and J. M. Hausdorff, "Toward automated, at-home assessment of mobility among patients with Parkinson disease, using a body-worn accelerometer," *Neurorehabil Neural Repair,* vol. 25, no. 9, pp. 810-8, Nov-Dec, 2011.

[4] J. J. Sosnoff, B. M. Sandroff, and R. W. Motl, "Quantifying gait abnormalities in persons with multiple sclerosis with minimal disability," *Gait Posture,* vol. 36, no. 1, pp. 154-6, May, 2012.

[5] T. A. Wren, S. Rethlefsen, and R. M. Kay, "Prevalence of specific gait abnormalities in children with cerebral palsy: influence of cerebral palsy subtype, age, and previous surgery," *J Pediatr Orthop,* vol. 25, no. 1, pp. 79-83, Jan-Feb, 2005.

[6] J. T. Zhang, A. C. Novak, B. Brouwer, and Q. Li, "Concurrent validation of Xsens MVN measurement of lower limb joint angular kinematics," *Physiol Meas,* vol. 34, no. 8, pp. N63-9, Aug, 2013.

[7] W. Zijlstra, and A. L. Hof, "Assessment of spatio-temporal gait parameters from trunk accelerations during human walking," *Gait Posture,* vol. 18, no. 2, pp. 1-10, Oct, 2003.

[8] A. Godfrey, S. Del Din, G. Barry, J. C. Mathers, and L. Rochester, "Instrumenting gait with an accelerometer: a system and algorithm examination," *Med Eng Phys,* vol. 37, no. 4, pp. 400-7, Apr, 2015.

[9] A. Hartmann, S. Luzi, K. Murer, R. A. de Bie, and E. D. de Bruin, "Concurrent validity of a trunk tri-axial accelerometer system for gait analysis in older adults," *Gait Posture,* vol. 29, no. 3, pp. 444-8, Apr, 2009.

[10] R. C. Gonzalez, A. M. Lopez, J. Rodriguez-Uria, D. Alvarez, and J. C. Alvarez, "Real-time gait event detection for normal subjects from lower trunk accelerations," *Gait Posture,* vol. 31, no. 3, pp. 322-5, Mar, 2010.

[11] R. Moe-Nilssen, and J. L. Helbostad, "Estimation of gait cycle characteristics by trunk accelerometry," *Journal of Biomechanics,* vol. 37, no. 1, pp. 121-126, 1//, 2004.

[12] M. Yang, H. Zheng, H. Wang, S. McClean, and D. Newell, "iGAIT: an interactive accelerometer based gait analysis system," *Comput Methods Programs Biomed,* vol. 108, no. 2, pp. 715-23, Nov, 2012.

[13] D. Kobsar, C. Olson, R. Paranjape, T. Hadjistavropoulos, and J. M. Barden, "Evaluation of age-related differences in the stride-to-stride fluctuations, regularity and symmetry of gait using a waist-mounted tri-axial accelerometer," *Gait & Posture,* vol. 39, no. 1, pp. 553-557, 1//, 2014.

[14] R. Saether, J. L. Helbostad, L. Adde, S. Braendvik, S. Lydersen, and T. Vik, "Gait characteristics in children and adolescents with cerebral palsy assessed with a trunk-worn accelerometer," *Res Dev Disabil,* vol. 35, no. 7, pp. 1773-81, Jul, 2014.

[15] J. K. Lee, and E. J. Park, "Quasi real-time gait event detection using shank-attached gyroscopes," *Med Biol Eng Comput,* vol. 49, no. 6, pp. 707-12, Jun, 2011.

[16] M. R. Patterson, and B. Caulfield, "Comparing adaptive algorithms to measure temporal gait parameters using lower body mounted inertial sensors," *Conf Proc IEEE Eng Med Biol Soc,* vol. 2012, pp. 4509-12, 2012.

[17] K. Aminian, B. Najafi, C. Bula, P. F. Leyvraz, and P. Robert, "Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes," *J Biomech,* vol. 35, no. 5, pp. 689-99, May, 2002.

[18] A. Salarian, H. Russmann, F. J. Vingerhoets, C. Dehollain, Y. Blanc, P. R. Burkhard, and K. Aminian, "Gait assessment in Parkinson's disease: toward an ambulatory system for long-term monitoring," *IEEE Trans Biomed Eng,* vol. 51, no. 8, pp. 1434-43, Aug, 2004.

[19] K. Bötzel, F. M. Marti, M. Á. C. Rodríguez, A. Plate, and A. O. Vicente, "Gait recording with inertial sensors – How to determine initial and terminal contact," *Journal of Biomechanics,* vol. 49, no. 3, pp. 332-337, 2/8/, 2016.

[20] B. R. Greene, D. McGrath, R. O'Neill, K. J. O'Donovan, A. Burns, and B. Caulfield, "An adaptive gyroscope-based algorithm for temporal gait analysis," *Med Biol Eng Comput,* vol. 48, no. 12, pp. 1251-60, Dec, 2010.

[21] P. H. Veltink, P. Slycke, J. Hemssems, R. Buschman, G. Bultstra, and H. Hermens, "Three dimensional inertial sensing of foot movements for automatic tuning of a two-channel implantable drop-foot stimulator," *Medical Engineering & Physics,* vol. 25, no. 1, pp. 21-28, 1//, 2003.

[22] B. Mariani, C. Hoskovec, S. Rochat, C. Bula, J. Penders, and K. Aminian, "3D gait assessment in young and elderly subjects using foot-worn inertial sensors," *J Biomech,* vol. 43, no. 15, pp. 2999-3006, Nov 16, 2010.

[23] M. Boutaayamou, C. Schwartz, J. Stamatakis, V. Denoel, D. Maquet, B. Forthomme, J. L. Croisier, B. Macq, J. G. Verly, G. Garraux, and O. Bruls, "Development and validation of an accelerometer-based method for quantifying gait events," *Med Eng Phys,* vol. 37, no. 2, pp. 226-32, Feb, 2015.

[24] K. Ben Mansour, N. Rezzoug, and P. Gorce, "Analysis of several methods and inertial sensors locations to assess gait parameters in able-bodied subjects," *Gait & Posture*.

[25] J. H. Hollman, M. K. Watkins, A. C. Imhoff, C. E. Braun, K. A. Akervik, and D. K. Ness, "A comparison of variability in spatiotemporal gait parameters between treadmill and overground walking conditions," *Gait Posture,* vol. 43, pp. 204-9, Jan, 2016.

[26] J. M. Jasiewicz, J. H. J. Allum, J. W. Middleton, A. Barriskill, P. Condie, B. Purcell, and R. C. T. Li, "Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals," *Gait & Posture,* vol. 24, no. 4, pp. 502-509, 12//, 2006.

[27] R. Moe-Nilssen, "A new method for evaluating motor control in gait under real-life environmental conditions. Part 1: The instrument," *Clin Biomech (Bristol, Avon),* vol. 13, no. 4-5, pp. 320-327, Jun, 1998.

[28] J. J. Kavanagh, and H. B. Menz, "Accelerometry: A technique for quantifying movement patterns during walking," *Gait & Posture,* vol. 28, no. 1, pp. 1-15, 7//, 2008.

[29] D. A. Winter, *Biomechanics and motor control of human movement*, Hoboken, N.J: John Wiley & Sons, 2009.

[30] O. Tirosh, and W. A. Sparrow, "Identifying heel contact and toe-off using forceplate thresholds with a range of digital-filter cutoff frequencies," *Journal of Applied Biomechanics,* vol. 19, no. 2, pp. 178-184, 2003.

[31] J. Martin Bland, and D. Altman, "STATISTICAL METHODS FOR ASSESSING AGREEMENT BETWEEN TWO METHODS OF CLINICAL MEASUREMENT," *The Lancet,* vol. 327, no. 8476, pp. 307-310, 2/8/, 1986.

[32] J. Lee Rodgers, and W. A. Nicewander, "Thirteen ways to look at the correlation coefficient," *The American Statistician,* vol. 42, no. 1, pp. 59-66, 1988.

[33] J. K. Taylor, and C. Cihon, *Statistical techniques for data analysis*: CRC Press, 2004.

[34] D. E. Hinkle, W. Wiersma, and S. G. Jurs, *Applied Statistics for the Behavioral Sciences*: Houghton Mifflin, 2003.

[35] H. B. Menz, S. R. Lord, and R. C. Fitzpatrick, "Acceleration patterns of the head and pelvis when walking on level and irregular surfaces," *Gait Posture,* vol. 18, no. 1, pp. 35-46, Aug, 2003.