2019 IEEE 16th International Conference on Rehabilitation Robotics (ICORR)
Toronto, Canada, June 24-28, 2019

Sensitivity comparison of inertial to optical motion capture during gait: implications for tracking recovery

Jeonghwan Lee, Sung Yul Shin, Gaurav Ghorpade, Tunc Akbas, and James Sulzer

Abstract— **Wearable sensors provide a foundation for development of wearable robotic technology to be used in clinical applications. Inertial motion capture (IMC) has emerged as a viable alternative to more cumbersome, non-portable optical methods. Previous work has validated the accuracy of IMC for gait compared to optical motion capture (OMC). However, it is unclear how well IMC can measure the small changes in gait function needed to gauge recovery. In this study, we evaluate the sensitivity of IMC compared to OMC to small changes in gait on a cohort of unimpaired individuals during treadmill walking. Eight individuals walked on a split-belt treadmill in three-minute trials with five randomized conditions: right belt speed decrementing at 0.05 m/s from 1.0 m/s, all with left belt held at 1.0 m/s, simulating recovery of hemiparetic gait. We extracted the root mean square deviation (RMSD) of joint kinematics between limbs and within the limb with modulated gait speed as the main outcome measure. We used linear mixed models to identify differences in sensitivity to changes in gait asymmetry and gait speed. Based on these models, we estimated the minimal detectible interval in gait parameters. We found that IMC was capable of measuring a difference in gait speed of 0.08 m/s, roughly the equivalent of two weeks recovery progress. Statistically we could not conclude a difference of sensitivity between IMC and OMC, although there is a strong trend that IMC is more sensitive to changes in gait. We conclude that IMC is a valid tool to measure progress in gait kinematics over the course of recovery.**

I. Introduction

Robotic gait therapy delivered by exoskeletons has become commercially available in recent years [1], [2]. One of the main advantages of robotic therapy is the ability to precisely measure delivered therapy and its effects. These measurements have been enabled by advancements in wearable sensor technology. For example, the current gold standard for measuring gait kinematics is optical motion capture (OMC) technology [3], [4]. With a lengthy setup time, small workspace, high-cost, as well as sensitivity to light, reflection and obstructions, OMC was not destined for implementation with mobile, wearable robotics. Up until recently, there was no viable alternative to portably measure 3D gait kinematics of the user.

Over the past decade, inertial motion capture (IMC), a portable, lower-cost alternative to OMC, has grown in popularity [5]–[9]. IMC is based on inertial measurement units (IMUs) composed of three-axis gyroscopes and accelerometers as well as a magnetometer. With the ability to transmit data wirelessly or store data over a period of days, IMC offers insight into previously inaccessible processes [6]. Further, IMC can typically be set up in a small fraction of the time required for setup of OMC, without the confounding issues of ambient lighting or visual occlusions. Recent studies have validated the use of IMUs to assess gait [10]–[15]. However, IMC has well-known drawbacks, primarily sensor drift and inhomogeneities in the ambient magnetic field that result in inaccuracies, typically approximately 1° (or more) of joint motion [16]. As a result, OMC remains the gold standard in motion capture due to its high accuracy (about 0.1 mm) [17], often a critical parameter in precision biomechanics analyses [18]. Yet such accuracy may not be required in many applications, including physical therapy. For example, perhaps the most important quantity to detect would be the *change* in gait function rather than the current state of the patient. Yet there is little information indicating how sensitive IMC is to small changes in walking performance.

The purpose of this work is to examine the sensitivity of IMC to small changes in gait in order to simulate its ability to track the recovery process after an injury such as stroke. We simulated the stroke recovery process by unilaterally varying the speed of one of the belts of a split-belt treadmill at discrete values. We were primarily interested in how well IMC could detect levels of symmetry between limbs based on these differences as well as changes within a limb. We then compared these results with simultaneously recorded OMC data. The implications of this work show that IMC is capable of tracking changes in gait function in a clinical setting as standalone sensors. IMC can be used in conjunction with robotic assistance or as an assessment tool to gauge progress.

List of abbreviations:
OMC – Optical motion capture
IMC – Inertial motion capture
IMU – Inertial measurement unit
RMSD – Root mean square deviation
EMDI – Estimated minimally detectible interval
CI – Confidence interval
MD – Mahalanobis distance

This work is supported by the Mission Connect Foundation Award 015-106

Jeonghwan Lee, Sung Yul Shin, Gaurav Ghorpade, Tunc Akbas, and James Sulzer are in the Walker Department of Mechanical Engineering at the University of Texas at Austin, Austin, TX 78712 USA. (phone: 512-471-0281; e-mail: james.sulzer@austin.utexas.edu).

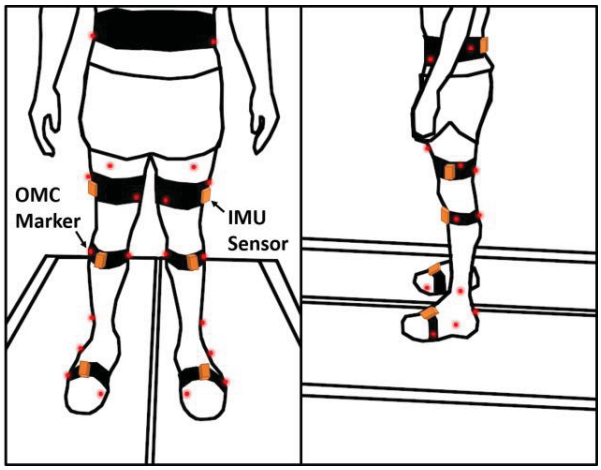

Figure 1. Optical motion capture (OMC) and inertial motion capture (IMC) marker setup shown on split-belt treadmill.

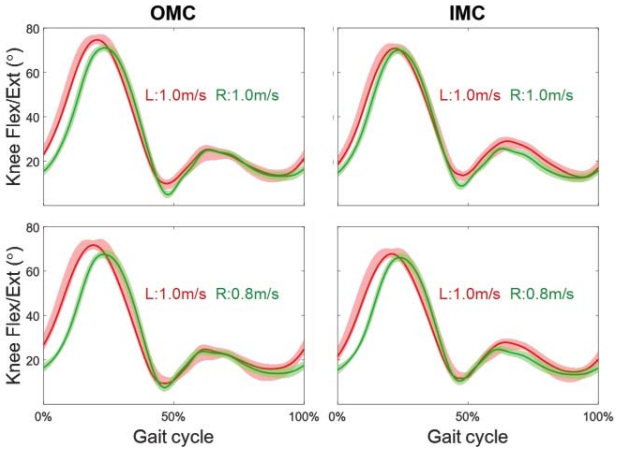

Figure 2. Knee joint angle information from one representative subject. Both OMC (left column) and IMC (right column) data shown with symmetric baseline (SYMM, top row) and maximum asymmetric (ASYMM, botttom row) conditions. Shaded areas indicate standard deviation. Note small changes to knee flexion in right side from visual inspection.

II. Data Acquisition

A. Participants and procedure

Gait kinematic data were collected from 7 males and 1 female, healthy subjects between ages of 21 and 33 years. All subjects were without neurological or cardiovascular conditions. No participants had a previous musculoskeletal injury in the lower limb within the last three months. This study was approved by The University of Texas at Austin Institutional Review Board and all participants provided written informed consent.

B. Experimental setup and protocols

A total of 13 lower limb joint motions were measured using both OMC and IMC systems simultaneously. We used PhaseSpace IMPROV (PhaseSpace Inc., San Leandro, CA, USA) for OMC and Xsens MVN (MVN Studio Version 4.3, Xsens Technologies B.V., Enschede, Netherlands) for IMC. The XSens system was selected specifically due to its proprietary robustness to noise and magnetic disturbance relative to other commercially available devices [19]–[21]. Each participant was outfitted with elastic velcro straps on which IMUs were placed on the pelvis, thighs, shanks and feet according the recommendations of the manufacturer. Active LED OMC markers were placed on each lower limb segment and pelvis as shown in Figure 1. Raw data was collected at 60 Hz for IMC and 240 Hz for OMC. Custom software was written in MATLAB R2016a (Mathworks, Inc., Natick, MA, USA) to post-process the raw data. The OMC data was resampled to 60 Hz and both IMC and OMC data were filtered with 4th order Butterworth filter at cut-off frequency of 10Hz [22].

Gait trials were performed on a split-belt treadmill (Bertec Inc., Columbus, OH, USA). In order to simulate hemiparetic walking impairment participants walked with one belt moving at a slower speed than the other. Participants walked at a baseline speed of 1.0 m/s on each belt, representative of healthy walking. This baseline trial lasted three minutes. We then exposed the participants to asymmetric treadmill walking at different speeds in pseudorandomized order (0.95 m/s, 0.90 m/s, 0.85 m/s and 0.8 m/s), all varying the left belt speed while keeping the right belt constant. The decrement in speed of 0.05 m/s was intended to simulate a small change in hemiparetic gait function. Each condition lasted three minutes, with one-minute breaks in between trials. All devices were re-calibrated after each trial to minimize systematic error. Due to technical errors in data collection, there was one missing asymmetry condition for four of the subjects.

III. Data Analysis

A. Data post-processing

Three-dimensional joint kinematics data from both IMC and OMC systems were extracted according to the recommendation of International Society of Biomechanics [23]. Total thirteen joint motions of interest including pelvis, hip, knee and ankle for both limbs were selected from the raw data. We first synchronized each trial data between OMC and IMC by using cross-correlation based on the reference of right hip flexion/extension. Next, the synched data was truncated into multiple single strides based on left heel strike events [24], and normalized into 101 frames. Processed data for all lower limb joint motions contained around 120 ~ 140 strides. For further analysis of sensitivity, 10 joint motions among all 13 joint motions were deployed: hip flexion/extension, hip abduction/adduction, hip internal/external rotation, knee flexion/extension, and ankle dorsi/plantar flexion for both limbs.

B. Sensitivity analysis

Both changes in gait symmetry and gait speed are important measures to characterize gait recovery [25], [26]. Thus, a sensitivity analysis was conducted along two different aspects, between-limb and within-limb joint kinematics, representing changes in gait asymmetry and gait speed, respectively. As a measure of asymmetry of gait, we used root mean square deviation (RMSD) between left and right lower limb joint motions. Since joint angles between lower limbs had a phase difference, we performed an additional synchronization procedure as described above. The extracted latencies from hip flexion/extension angles of each stride were applied across all other joint motions. Then, RMSDs of each synchronous pair were calculated. The example of synchronized joint tra-

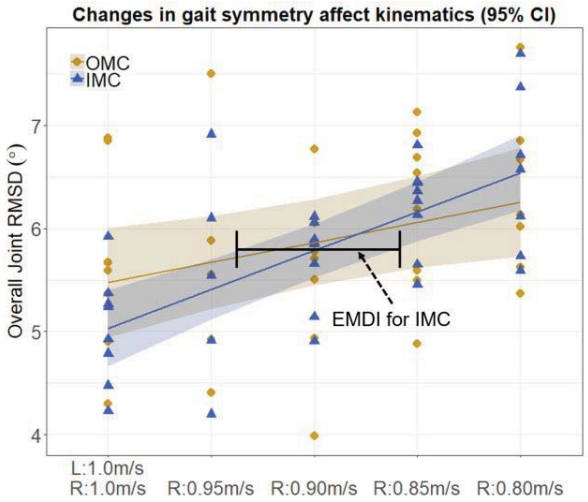

Figure 3. Estimated minimally detectible interval (EMDI) illustrated on the RMSD of overall joint motion for both OMC and IMC. The EMDI is calculated by the minimum horizontal distance stretching from the upper 95% confidence interval (CI) to the lower 95% CI. While EMDI exists for IMC, it does not exist for OMC because the relatively low sensitivity (slope) and large CI.

jectories for one subject is shown in Figure 2. We also examined within-limb changes in joint kinematics induced by changes in gait speed. We extracted the RMSDs of within-limb joint kinematics between the baseline condition (1.0 m/s) and other right tread speeds.

For statistical analysis, 10 continuous strides were sampled at three different portions of an individual trial, that is, within the 2nd, 3rd and 4th quarter of a trial. The first quarter was excluded to account for adaptation to the novel imposed gait pattern. Then, the average RMSD of 30 strides was calculated as a dependent variable for each joint. To obtain a holistic view of all joint motions, we calculated the Mahalanobis distance (MD), a well-known method of combining data of different dimensions [27]. The Mahalanobis distance accounts for the covariance of the data to standardize the scaling of the distance measure in multi-dimensional space. Thus, MD enables to combine RMSDs of multiple joints in a fair manner.

We evaluated sensitivity to walking perturbations in IMC and OMC systems using a linear mixed effects model based on two independent variables, system type (IMC and OMC) and symmetry condition (symmetric and four additional asymmetric conditions). The model is described in (1), where $\mathbf{y}$ is the response vector, β_I is the intercept, $\mathbf{X_C}$ is the vector of predictor variables for conditions, β_C is the fixed effect regression coefficient for conditions, $\mathbf{X_M}$ is the vector of predictor variables for system modality and β_M is the fixed effect regression coefficient for system modality, $\mathbf{Z_S}$ is the vector of the random effect for subjects, u_S is the random effect coefficients for subjects and $\boldsymbol{\varepsilon}$ is the vector of the residuals for observations.

$$\mathbf{y} = \beta_I + \beta_C\mathbf{X_C} + \beta_M\mathbf{X_M} + u_S\mathbf{Z_S} + \boldsymbol{\varepsilon}. \quad (1)$$

All statistical comparisons were conducted using R version 3.52 (The R Foundation, 2018). We used this model to answer two separate questions: first, how sensitive were IMC and OMC to changes in gait asymmetry, and second, how sensitive these systems were to changes in gait speed. Our primary measure of sensitivity was β_C. Our secondary outcome measure was the difference in measured motion between the two systems, a reflection of the overall accuracy assuming that OMC is the gold standard. This measure of accuracy was β_M.

From the same linear mixed effects models, we extracted the estimated minimally detectible interval (EMDI) of both in gait asymmetry and speed. This quantity provides a more ecological view of what the data represents because it estimates the reliable range of minimum change in gait speed (or symmetry) that can be detected by the motion capture systems. The EMDI is the horizontal distance between the upper 95% confidence interval (CI) and the lower 95% CI of the linear regression as illustrated in Figure 3. Often there was no horizontal distance that connected these two boundary conditions due to low β_C and/or high variance. The EMDI thus represents a clinically relevant measure of what change in gait function can be measured using motion capture.

IV. Results

A. Sensitivity to Change in Gait Asymmetry

Both systems showed a relationship with changes in gait symmetry. Based on overall joint motion (calculated from MD), we found a significant change with increasing asymmetry in OMC ($\beta_C = 0.20$, $p<0.05$) as well as with IMC ($\beta_C = 0.38$, $p<0.001$). There was no observed difference in overall measured motion between the systems ($p=0.56$). We did not observe a significant change in asymmetry within the hip for either system, with the exception of hip internal/external rotation in the IMC system ($\beta_C = 0.33$, $p<0.05$). We observed a difference in measured motion in hip flexion/extension of 1.12° ($p<0.001$) between the two systems. We found that IMC measurement of knee flexion/extension varied with changing asymmetry ($\beta_C = 0.51$, $p<0.01$), but no observed difference in measurement between systems ($p=0.07$). Ankle dorsi/plantarflexion measurements in both the OMC ($\beta_C = 0.31$, $p<0.01$) and IMC ($\beta_C = 0.64$, $p<0.001$) systems correlated with changes in asymmetry, with no observed difference between systems ($p=0.29$). There were also no differences in sensitivity between the motion capture systems, as all confidence intervals overlapped. Results are summarized in Table I.

B. Sensitivity to Change in Gait Speed

We also investigated the sensitivity to changes in gait speed within the right (varied) limb. MD showed changes in walking kinematics in both OMC ($\beta_C = 0.42°$, $p<0.001$) and IMC ($\beta_C = 0.35$, $p<0.001$) systems, with a mean difference between systems of 0.53°. We observed changes in walking kinematics in all joints with varying speed in both OMC and IMC. There were observed difference between two systems in hip abduction/adduction of 0.32° ($p<0.05$) and knee extension/extension of 0.86° ($p<0.01$). We did not observe notable difference between the motion capture systems in other joints. Results are summarized in Table II.

C. Estimated Minimally Detectible Interval

Table III shows EMDI for changes in gait symmetry and gait speed in both systems. For gait symmetry, we found EMDIs in IMC for overall joint motion (0.07 m/s) and ankle dorsi/plantar flexion (0.12 m/s). No EMDIs were found for OMC for changes in gait symmetry. For changes in gait speed,

TABLE I. SENSITIVITY TO CHANGE IN GAIT ASYMMETRY

Joint Angles RMSD (°)		OMC β_C [95% CI]	IMC β_C [95% CI]	OMC - IMC β_M [95% CI]
Overall (MD)		0.20 * [0.02, 0.37]	0.38 *** [0.24, 0.51]	-0.09 [-0.40, 0.22]
Hip	Flex/Ext	0.05 [-0.22, 0.32]	0.12 [-0.10, 0.34]	-1.12 *** [-1.70, -0.54]
	Abd/Add	0.04 [-0.24, 0.31]	0.11 [-0.09, 0.32]	-0.28 [-0.73, 0.16]
	Int/Ext Rot	0.28 [-0.05, 0.62]	0.33 * [0.01, 0.66]	0.32 [-0.28, 0.94]
Knee Flex/Ext		0.25 [-0.16, 0.67]	0.51 ** [0.20, 0.82]	-0.66 [-1.38, 0.05]
Ankle Dorsi/Plantar		0.31 ** [0.09, 0.53]	0.64 *** [0.45, 0.83]	-0.23 [-0.67, 0.20]

***p< 0.001, **p< 0.01, *p< 0.05

TABLE II. SENSITIVIY TO CHAGE IN GAIT SPEED

Right Leg Joint Angles RMSD (°)		OMC β_C [95% CI]	IMC β_C [95% CI]	OMC - IMC β_M [95% CI]
Overall (MD)		0.42 *** [0.23, 0.61]	0.35 *** [0.22, 0.48]	-0.53 ** [-0.85, -0.21]
Hip	Flex/Ext	0.54 * [0.09, 0.99]	0.66 ** [0.18, 1.14]	-0.26 [-1.11, 0.60]
	Abd/Add	0.23 ** [0.07, 0.40]	0.52 *** [0.26, 1.04]	-0.32 * [-0.64, 0.00]
	Int/Ext Rot	0.54 ** [0.09, 0.99]	0.57 * [0.12, 1.02]	0.11 [-0.71, 0.93]
Knee Flex/Ext		0.42 *** [0.11, 0.73]	0.41 *** [0.19, 0.62]	-0.86 ** [-1.42, -0.29]
Ankle Dorsi/Plantar		0.55 *** [0.36, 0.74]	0.64 *** [0.45, 0.83]	-0.04 [-0.38, 0.31]

***p< 0.001, **p< 0.01, *p< 0.05

TABLE III. ESTIMATED MINIMALLY DETECTIBLE INTERVAL

EMDI (m/s)		Asymmetry OMC	Asymmetry IMC	Gait Speed OMC	Gait Speed IMC
Overall (MD)		-	0.07	0.10	0.08
Hip	Flex/Ext	-	-	-	-
	Abd/Add	-	-	-	0.16
	Int/Ext Rot	-	-	-	-
Knee Flex/Ext		-	-	-	0.12
Ankle Dorsi/Plantar		-	0.12	0.11	0.10

Not Available "-"

EMDI were found for IMC in overall joint motion (0.08 m/s, hip abduction/adduction (0.16 m/s), knee flexion/extension (0.12 m/s) and ankle dorsi/plantar flexion (0.10 m/s), whereas EMDI were found for OMC only for overall (0.10 m/s) and ankle dorsi/plantar flexion (0.11 m/s).

V. DISCUSSION

The purpose of this paper is to compare the sensitivity of IMC and OMC to small changes in gait simulating the recovery process after stroke. We observed that both systems were sensitive in overall kinematics to changes in gait symmetry and gait speed, and to some extent, at the individual joint level. Thus, we suggest that IMC can be an effective measurement tool for monitoring recovery following stroke.

We found that IMC was sensitive enough to detect changes in gait symmetry reflecting recovery after stroke. These changes were observed strongly in overall kinematics and in sagittal plane knee and ankle motion. OMC data also showed similar trends, except for the knee flexion/extension. A comparison of regression values between systems showed no statistically observable difference in sensitivity. However, EMDI values were lower for IMC than OMC, indicating greater sensitivity with IMC (Table III).

We also found comparatively greater sensitivity to changes in gait speed in both systems compared to the symmetry analysis. Sensitivity measures were mostly higher and EMDI were observed more often in the case of changes in gait speed compared to gait symmetry. Significant differences with increased belt speeds were observed for both modalities in every joint and overall kinematics. These results suggest that IMC is capable of detecting small changes in gait speed, even at an individual joint level. It is unclear what accounts for this trend, but this could result from greater variability in the left leg during asymmetric walking patterns.

Previous work has examined the sensitivity of IMUs to changes in spatiotemporal symmetry compared to the gait parameters measured by an instrumented treadmill as ground truth [14]. Varying the split-belt speed by increments of 0.25 m/s, these researchers found that IMC (APDM Opal, APDM Inc., Portland, OR, USA) was capable of measuring this difference. In contrast, our results showed that joint kinematic information over the lower limbs can provide a more sensitive measure of changes in gait symmetry and gait speed. We were able to detect differences of 0.08 m/s, three times less than the interval tested in [14].

The difference between 0.08 m/s and 0.25 m/s is clinically important. Our recent work measured gait kinematics longitudinally using IMC during the recovery process in the acute phase (Week 1 to Week 12) in mild to moderately impaired individuals following stroke [28]. We observed that the change in gait speed per week of recovery was roughly equivalent to a change in belt speed of 0.034 m/s. Taken together, this would mean that the use of joint kinematic information from IMC is at least sensitive enough to detect a change in gait speed just over two weeks of the recovery process. It should be noted that while a linear model of recovery is over-simplistic, this estimate is meant to provide a rough idea of clinical performance.

We found small differences in measurement between IMC and OMC consistent with previous literature. Previous validation studies have reported that the accuracy of IMC in dynamic motion is between 1.9 ~ 3.5° [19], thus we expected to find similar differences in our measurements. For example, we noted a significant difference in hip flexion of 1.12° in gait symmetry analysis (Table I), whereas we did not observe a difference in changes in gait speed (Table II). This could be due to the variability in both systems, given the confidence

intervals within 2° in all comparisons. Overall our findings were consistent with earlier work in accuracy.

The data provided in this study offer a limited perspective of sensitivity. For example, our sensitivity analysis could be improved with more sophisticated techniques. A larger pool of participants would likely result in greater sensitivity estimates by reducing the 95% CI. Recording individuals on multiple days could have also improved robustness of data. Our results are limited to the Xsens MVN system and could differ with other IMC systems. As such, we cannot conclusively state the maximum sensitivity of IMC, but we can offer an initial estimate. Regardless, it appears that IMC is capable of being as sensitive to changes in gait as OMC.

VI. Conclusion

Our results show that inertial motion capture is a valid tool for measuring clinically relevant changes in gait kinematics reflecting recovery after stroke. The performance is as sensitive as the current standard in kinematic tracking, optical motion capture. Together with robotic technology, inertial measurement units can quantify therapy delivery and progress, improving robustness of outcomes.

References

[1] I. Díaz, J. J. Gil, and E. Sánchez, "Lower-Limb Robotic Rehabilitation: Literature Review and Challenges," *J. Robot.*, vol. 2011, pp. 1–11, Nov. 2011.

[2] J. Mehrholz and M. Pohl, "Electromechanical-assisted gait training after stroke: A systematic review comparing end-effector and exoskeleton devices," *Journal of Rehabilitation Medicine*, vol. 44, no. 3. pp. 193–199, 2012.

[3] M. W. Whittle, "Gait analysis," in *The Soft Tissues*, Elsevier, 1993, pp. 187–199.

[4] S. R. Simon, "Quantification of human motion: gait analysis—benefits and limitations to its application to clinical problems," *J. Biomech.*, vol. 37, no. 12, pp. 1869–1880, Dec. 2004.

[5] W. Tao, T. Liu, R. Zheng, and H. Feng, "Gait analysis using wearable sensors," *Sensors*, vol. 12, no. 2. Molecular Diversity Preservation International, pp. 2255–2283, 16-Feb-2012.

[6] P. B. Shull, W. Jirattigalachote, M. A. Hunt, M. R. Cutkosky, and S. L. Delp, "Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention," *Gait and Posture*, vol. 40, no. 1. Elsevier, pp. 11–19, 01-May-2014.

[7] A. Muro-de-la-Herran, B. García-Zapirain, and A. Méndez-Zorrilla, "Gait analysis methods: An overview of wearable and non-wearable systems, highlighting clinical applications," *Sensors (Switzerland)*, vol. 14, no. 2. Multidisciplinary Digital Publishing Institute, pp. 3362–3394, 19-Feb-2014.

[8] S. Sprager and M. B. Juric, "Inertial sensor-based gait recognition: A review," *Sensors (Switzerland)*, vol. 15, no. 9. Multidisciplinary Digital Publishing Institute, pp. 22089–22127, 02-Sep-2015.

[9] I. H. Lopez-Nava and A. Munoz-Melendez, "Wearable Inertial Sensors for Human Motion Analysis: A Review," *IEEE Sens. J.*, vol. 16, no. 22, pp. 7821–7834, 2016.

[10] T. Cloete, "Benchmarking full-body inertial motion capture for clinical gait analysis Benchmarking full-body inertial motion capture for clinical gait analysis," *Conf. Proc. Int. Conf. IEEE Eng. Med. Biol. Soc.*, vol. 2008, no. January, pp. 4579–82, Aug. 2009.

[11] T. Cloete and C. Scheffer, "Repeatability of an off-the-shelf, full body inertial motion capture system during clinical gait analysis," in *2010 Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBC'10*, 2010, pp. 5125–5128.

[12] 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, p. N63, Jul. 2013.

[13] C. Nüesch, E. Roos, G. Pagenstert, and A. Mündermann, "Measuring joint kinematics of treadmill walking and running: Comparison between an inertial sensor based system and a camera-based system," *J. Biomech.*, vol. 57, pp. 32–38, May 2017.

[14] E. P. Washabaugh, T. Kalyanaraman, P. G. Adamczyk, E. S. Claflin, and C. Krishnan, "Validity and repeatability of inertial measurement units for measuring gait parameters," *Gait Posture*, vol. 55, pp. 87–93, Jun. 2017.

[15] M. Al-Amri, K. Nicholas, K. Button, V. Sparkes, L. Sheeran, and J. Davies, "Inertial Measurement Units for Clinical Movement Analysis: Reliability and Concurrent Validity," *Sensors*, vol. 18, no. 3, p. 719, Feb. 2018.

[16] G. Ligorio and A. M. Sabatini, "Dealing with magnetic disturbances in human motion capture: A survey of techniques," *Micromachines*, vol. 7, no. 3, p. 43, Mar. 2016.

[17] P. Eichelberger *et al.*, "Analysis of accuracy in optical motion capture - A protocol for laboratory setup evaluation.," *J. Biomech.*, vol. 49, no. 10, pp. 2085–2088, Jul. 2016.

[18] T. Akbas, R. R. Neptune, and J. Sulzer, "Neuromusculoskeletal Simulation Reveals Abnormal Rectus Femoris-Gluteus Medius Coupling in Post-stroke Gait," *Front. Neurol.*, vol. 10, p. 301, 2019.

[19] A. Godwin, M. Agnew, and J. Stevenson, "Accuracy of Inertial Motion Sensors in Static, Quasistatic, and Complex Dynamic Motion," *J. Biomech. Eng.*, vol. 131, no. 11, p. 114501, Nov. 2009.

[20] K. Lebel, P. Boissy, M. Hamel, and C. Duval, "Inertial measures of motion for clinical biomechanics: Comparative assessment of accuracy under controlled conditions - Effect of velocity," *PLoS One*, vol. 8, no. 11, p. e79945, Nov. 2013.

[21] K. Lebel, P. Boissy, M. Hamel, and C. Duval, "Inertial measures of motion for clinical biomechanics: Comparative assessment of accuracy under controlled conditions - Changes in accuracy over time," *PLoS One*, vol. 10, no. 3, p. e0118361, Mar. 2015.

[22] J. Sinclair, P. John Taylor, and S. Jane Hobbs, "Digital filtering of three-dimensional lower extremity kinematics: An assessment," *J. Hum. Kinet.*, vol. 39, no. 1, pp. 25–36, 2013.

[23] G. Wu *et al.*, "ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion - Part I: Ankle, hip, and spine," *J. Biomech.*, vol. 35, no. 4, pp. 543–548, Apr. 2002.

[24] J. A. Zeni, J. G. Richards, and J. S. Higginson, "Two simple methods for determining gait events during treadmill and overground walking using kinematic data," *Gait Posture*, vol. 27, no. 4, pp. 710–714, May 2008.

[25] L. N. Awad, J. A. Palmer, R. T. Pohlig, S. A. Binder-Macleod, and D. S. Reisman, "Walking speed and step length asymmetry modify the energy cost of walking after stroke," *Neurorehabil. Neural Repair*, vol. 29, no. 5, pp. 416–423, Jun. 2015.

[26] B. Balaban and F. Tok, "Gait Disturbances in Patients With Stroke," *PM and R*, vol. 6, no. 7. No longer published by Elsevier, pp. 635–642, 01-Jul-2014.

[27] R. De Maesschalck, D. Jouan-Rimbaud, and D. L. Massart, "The Mahalanobis distance," *Chemom. Intell. Lab. Syst.*, vol. 50, no. 1, pp. 1–18, Jan. 2000.

[28] S. Y. Shin, R. K. Lee, P. Spicer, and J. Sulzer, “Quantifying dosage of physical therapy using lower body kinematics: a longitudinal pilot study on early post-stroke individuals,” no. In review, 2019.